Intention detection based exoskeleton for gait neurorehabilitation

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Abstract

The prevalence of neurological diseases such as stroke, spinal cord injury and traumatic brain injury is increasing quickly in the industrialised societies. Although the benefit of the use of technology in rehabilitation and neurorehabilitation programs is proved, the presence of mechatronic systems is still very low.

This paper proposes a new lower limb exoskeleton for gait rehabilitation in persons with neurological pathologies. Since potential users have very reduced mobility even to start common daily movements, the control of the exoskeleton has to be intention based. The estimation of the intention of the user is based on hip and knee angle, and the EMG signal is monitored for intention detection, control and neurofeedback aims.

A novel approach of the EMG to force conversion in paraplegic patients is also described.

1. Introduction

The prevalence of other neurological diseases such as spinal cord injury, stroke, traumatic brain injury is increasing quickly in the industrialised societies. Population ageing, the stressing way of life or new habits are some of the factors causing this increase. Additionally, the improvements introduced in the protocols of immediate assistance are increasing the survival rate in the accidents and as a result, the prevalence of the spinal cord injury for example has been doubled in the last 20 years. The prevalence and incidence of stroke, spinal cord injury and traumatic brain injury statistics are shown in the Table 1.

	Prevalence	**Incidence**
Stroke	1690	220
Spinal cord injury	900	2.5
Traumatic brain injury	200	175-200

Table 1: stroke, spinal cord injury and traumatic brain injury statistics in x per 100.000 habitants. Data obtained from de American DHSS.

Neuromuscular diseases cause mobility impairments and depending on the level of affectation, gait may be affected drastically making walking even impossible. Rehabilitation process is critical for recovering global mobility and motivation, intensity and rehabilitation quality are key factors for successful rehabilitation.

The global objective of the work presented in this paper is the development of a lower limb exoskeleton for gait rehabilitation in people suffering from neurological disorders, compensating the patient's functional activity. Nowadays the lower limb and gait rehabilitation procedures are mainly based on manual physiotherapy and specific muscular exercises. The use of the robot mediated therapy improves the actual rehabilitation procedures, allowing setting up a more effective rehabilitation, gaining motivation for the patient, independence from availability of occupational therapists, gaining repeatability, precision and motivation, and allowing the patient developing gait patterns from the early stages of the process.

In the last years several initiatives for the developments of robotic platforms for gait rehabilitation have been launched, but few have been successful and none of them is portable, reducing very much the possible applications.

Regarding this, Fatronik is developing a portable, user transparent solution for gait rehabilitation, to encourage the user to interact with the exoskeleton and avoid passive motion of the body-machine structure. The portability of this solution will allow also developing rehabilitation activities in different scenarios.

The developed exoskeleton is not targeting specific pathology, but patients that, as consequence as different pathologies, have balance, basic mobility and muscular coordination problems. However, it's necessary to estimate an initial estimation of the level of support to be applied by exoskeleton, in order to determine the specifications of the necessary drives. As first hypothesis, it has been estimated that users are able to keep on one's feet, being able to give 50% of the required torque during gait. This value will have to be reconsidered after experimental tests.

This paper presents the first year development of the described exoskeleton. First of all, a state of art analysis is presented, giving an idea of the actually existing lower limb exoskeletons. Secondly, the development of the mechanical design (actuation system and the structural characteristics) and the control design is exposed. with the papers includes conclusions of the developments and further work to be done.

2. State of the art

Nowadays there are various types of exoskeletons at research level. The most notable are the Hybrid Assistive Leg (HAL), developed by Prof. Sankai at the Cybernics Laboratory of the University of Tsukuba in Japan [1] and the Berkeley Lower Extremity Exoskeleton (BLEEX), developed by Prof. Kazerooni at the University of Berkeley in the USA.

The power assist device HAL is a walking aid system for people with walking disorders. It successfully walks and carries its own power supply and has been designed to assist the wearer's muscles by measuring users own muscle activity. The HAL team proposes a control method using biological and motion information, thus the exoskeleton produces a torque depending on the control strategy helping the user's motion in a different way. The model of user's lower limb equipped with HAL was constructed in order to estimate operator's viscoelasticity by using the impedance control method. The first HAL prototype had only two degrees of freedom actuated with electrical motors, the hip flexion-extension and the knee flexion-extension. In the last version of HAL, the ankle dorsi-plantar flexion has been driven too.

BLEEX is a robotic exoskeleton for human performance augmentation capable of carrying its own weight plus an external payload [2]. It's energetically autonomous and walks at the average speed of 1.3 m/s while carrying a 34 kg payload. The BLEEX control scheme is solely based on measurements from the exoskeleton, so there are no direct measurements from the user or from the human-machine contact areas [3] and the problems associated with measuring interaction forces or human muscle activity disappear. Concerning on the actuated degrees of freedom (dof), this exoskeleton has 7 dof driven with hydraulic cylinders: 3 at the hip, 1 at the knee and 3 at the ankle.

Additionally, other minor initiatives have also been launched, as the Roboknee, Lokomat and Lopes devices.

The RoboKnee is a powered knee brace that works in parallel to the wearer's knee but does not transfer loads to the ground. This device transfers the weight of the load to the human skeleton. The particularity of the RoboKnee is the mechanical contribution to achieve a high level of transparency. Low impedance is reached through the use of a Series Elastic Actuators (SEA). In this exoskeleton the user intent is determined through the ground reaction forces and the knee joint angle [4].

Roboknee is a powered knee brace that works in parallel to the wearer's knee without transferring loads to the ground. The particularity of the Roboknee is the mechanical contribution to achieve a high level of transparency, without interference between the exoskeleton frame and the user. The knee flexion-extension is actuated by a Serie Elastic Actuator, that permits achieving a low mechanical impedance. In this device the user intent is determined through the knee joint angle and ground reaction forces.

The Lokomat and the LOPES are gait rehabilitation robots. The Lokomat is a four dof robotic orthosis [5] which proposes adaptive control methods that minimize the interaction forces with the patient with respect to an adaptable reference pattern, controlling the entire gait cycle. In opposition, the LOPES exoskeleton aims to support and not take over those tasks that the patient is unable to perform without help, using an impedance control scheme [6].

3.Technical development

The technical development of the exoskeleton has been divided in two different lines; the mechanical development and the development of the control system, including the intent detection.

The mechanical development is based in the biomechanical analysis of the human morphology and the human gait, since the exoskeleton must fit the patient in an ergonomic way and must be able to assist patients during gait movements. This includes not only mechanical structure but also selection of drives. In order to have a preliminary exoskeleton to perform experimental analysis, a first complete lower limb prototype have been designed.

The control system of the exoskeleton includes the method to detect user intention and the decision making to assist patient properly during gait with the required torque in each joint. In order to achieve a robust system, the control system has been development modularly. Concerning this a preliminary one-joint (knee) prototype has been produced in order to test and validate the control system. Based on the results of this single-joint prototype, the development of whole lower limb exoskeleton control system has been launched.

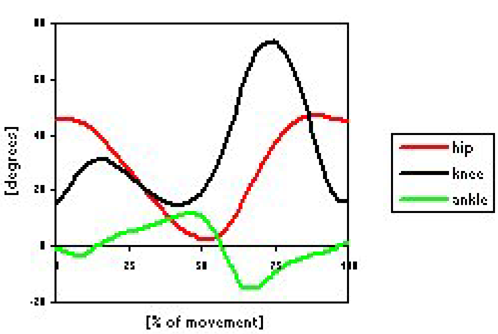
3. Mechanical design

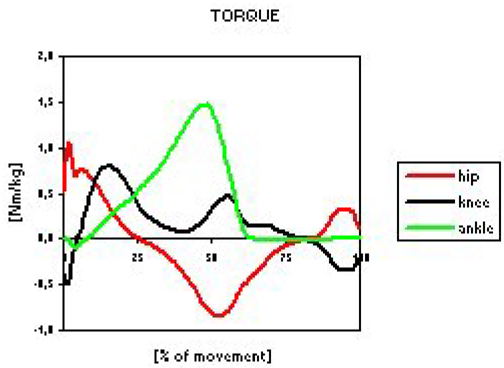
As explained before, the mechanical development of the exoskeleton includes not only structural design but also drive selection. Actually, selection of proper drives is challenging since human joints require high torques during gait but t the same time, aesthetic issues requires compact and low weight drives.

3.1. Drive design

To determine which degrees of freedom were have to be actuated in the exoskeleton, the most common daily live activities have been analysed: gait cycle, sitting down and standing up movements, and going up and downstairs activities.

The angle pattern data of the different lower limb joints in the sagital plane during the gait cycle, and the moments and powers of this plane measured in this movement are shown in Figure 1. This data of the gait cycle have been obtained from different normalised gait experiments [7]. These experiments are developed in normal conditions by regular size users and at normalised speed. Even this data depends on users involved in experiments (anthropomorphic data is different in different countries), it gives a enough precise base for a biomechanical estimation. ..





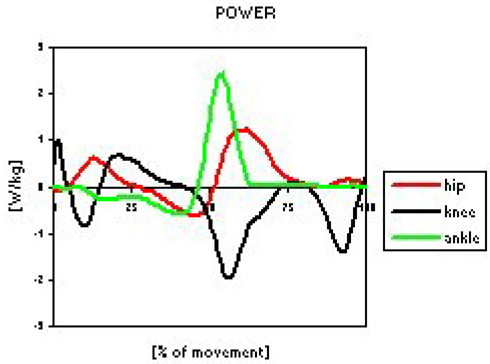


Figure 1: Angle, moment and power pattern values in the sagital plane during the gait cycle.

The maximum power consumption of each joint is shown in Table 2.

The values were computed considering a patient weight of 75 kg wearing a 30 kg weight exoskeleton, being the patient able to give 50% of the required torque. As explained in [8] the most torque required movements in any daily live activity, are in the sagital plane. Therefore, the mechanically actuated degrees of freedom are three: the hip flexion-extension, the knee flexion-extension and the ankle dorsal-plantar flexion. In this first mechanical prototype the hip flexion-extension and knee flexion-extension are driven by motors, and the ankle dorsal-plantar flexion is driven by a spring that brings the foot to its natural position during the swing phase to avoid dragging the feet.

		Gait	Stairs up	Stairs down	Power max (W)
	Flexion	82,55	**87,75**	33,75	
	Extension	-42,32	0	-114,75	87,75
	Abduction	26,19			
	Adduction	-25,52			26,19
	Ext. Rotation	4,93			
Hip	Int. Rotation	-2,16			4,93
Knee	Flexion	67,37	**195,75**	13,5	
	Extension	-131,3	-20,25	-324	195,75
	Varus				
	Valgus				
	Ext. Rotation				

	Int. Rotation				
Ankle	Dors. flexion	165,0	**195,75**	67,5	195,75
	Plant. flexion	-39,56	-13,5	-249,75	
	Eversion	2,97			2,97
	Inversion	-0,81			
	Extension				
	Flexion				

Table 2: Maximum power consumption values of the lower limb joints.

The specifications of each actuated dof (Table 3) have been determined to select the most appropriate drives for the exoskeleton.

	Angle (º)	Torque (Nm)	Power (W)	Velocity (rpm)
HIP	-10 / 75	72	87,75	25,78
KNEE	3,2 / 100	101,25	195,75	45,83

Table 3: Angle, torque, power and velocity specifications of the actuated joints

The volume, the weight an the aesthetic also have to be considered in the drive selection. Three different actuators come into question: hydraulic cylinders, pneumatic artificial muscles and electrical motors.

The easiest way to use hydraulic cylinders consists on situating one of the cylinders transversally in each joint. As the patient will have more difficulties to sit down, the cylinders have to be placed in another configuration with the subsequent angle restriction. In addition, this actuation system needs a hydraulic tank, a pumping system and electrical batteries to work properly. Concerning the weight of a hydraulic system, the oil tank would weigh about 3 kg, the pump would weight around 1 kg and the system needs a 100-150 bar pressure circuit, which could make difficult the design of the user's security. Moreover, due to the high torque specifications of the joints, the cylinders should have an external diameter greater than 30 mm and it would be necessary to add a position measurement sensor and a micro valve in each cylinder. Once all additional systems have been determined, a total weight of the hydraulic system (without batteries, electronic and control) of 20 kg minimum has been estimated. As a result, this alternative to actuate the exoskeleton with an hydraulic system has been ruled out.

Another option is the use of pneumatic artificial muscles (PAM) as actuators, due to their lightness, flexibility and their capacity to produce a high amount of force. To reach the specifications needed in each actuated joint, the muscles would have a diameter greater than 40 mm and a weight of 800 g. Moreover, as they are simple effect actuators, two muscles are necessary to actuate each dof, which means 1600 g considering only the PAMs without any other additional system. The main disadvantages of using this actuation system are based in the additional elements needed, especially in the compressor. It provides air with an appropriate pressure to the muscles, but it has an internal high pressure with the consequent security risk and its weight would exceed 3 or 4 kg. Therefore, it's necessary to find a lighter option with better appearance.

The third option is using electrical motors. This option would simplify the design and the aesthetic effect is considerably lower than with the other options. Other important advantages of this solution are the small volume and weight, and the absence of any pressure system and any additional system apart from the batteries. Moreover, including the electrical motors in the exoskeleton wouldn't imply any angle restriction in the joints and they could be restricted electrically or mechanically.

In this kind of actuation, the high torque requirement in each joint is a problem that needs to be solved. After having analysed the commercial solutions of the servomotors and reduction gears, as an assembly electrical motor-reduction gear with a small volume and weight but with necessary output torque, speed and power hasn't been found. A drive including a Robodrive motor kit (rotor and stator) and a Harmonic Drive compact reduction gear (Figure 2) reaching the objective values has been designed. The estimated output torque speed and power are shown in Table 4.

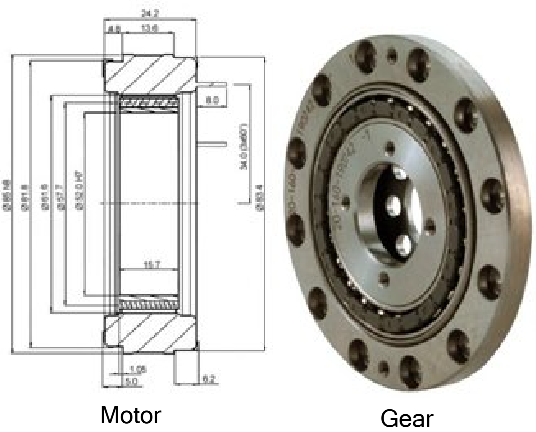


Figure 2: Selected Robodrive motor kit and Harmonic Drive reduction gear.

	Nom. Torque (Nm)	Max. Torque (Nm)	Speed (rpm)	Power (W)
Motor 85x13 Gear: CSD 32	71,5	151	60	450

Table 4: Torque, speed and power values achieved with the servomotor and reduction gear selected.

Finally the designed drive has a diameter of 113 mm, a 104.5 mm length at weighs 2,8 kg.

3.2. Structural characteristics

An essential aim is to design a suitable exoskeleton for most people possible. Depending on sex, weight, height and age of the patient, the anthropometric measures change considerably. The exoskeleton has been designed to cover the different physiognomies of people in a height range from 1550 mm to 1850 mm: the thigh is adjustable from 380 to 470 mm and the calf from 360 to 450 mm [9]. Another aspect to be considered is the easiness of the exoskeleton's regulation. Therefore, a quick adjustment method with some simple butterfly screws has been included.

When designing the mechanical structure of the exoskeleton it important to set the range of motion in each joint. The exoskeleton must allow free movements to patients within the ranges of motion required during gait or sitting down, but at the same time the exoskeleton can't go farther than maximum range of movement of patient in each joint in order to avoid any damage. Regarding this the range of motion in each joint are a bit smaller than values allowed by the biomechanical joints to avoid any kind of injuries. The angle range allowed in the exoskeleton is shown in Table 5.

The angle convention was set as zero in quiet standing trial; positive values mean ankle dorsi flexion, knee and hip flexion, negative values mean ankle plantar flexion, knee and hip extension.

	Angle (º)
HIP	115/-12
KNEE	0 / -115

Table 5: Angle range of the actuated joints allowed in the exoskeleton.

It's especially important to assure that the knee joint position doesn't exceed an angle of 0º. Apart from the mechanical stops that used in each dof, there are some electrical stops situated some degrees before the mechanical ones. In case an electrical problem occurred, the mechanical stops would act.

Another important aspect to consider in the structural design process is the comfort of the exoskeleton for the patient. Therefore, the structural elements in contact with the user (hip, thigh, calf and insole) have been designed with the advice of the orthopaedics specialist J. M. Vallecillo, using materials and shapes that assure the patient comfort but with a rigid enough to guarantee the transmission of the movement to the patient.

Finally, to validate the structural design, FE analyses have been made to assure that the structure resists the applied loads. As these forces change during the movements, the studied cases were the most unfavourable cases along the gait cycle: the points of maximal joint torques and ground reaction forces. A human body model was included to represent as realistic as possible, the human-exoskeleton interaction. After the stress and strain distributions in the structure were identified, the weakest area corresponds to the steel sheet situated between the ankle and the foot. The maximum stress obtained in this area is about 60 MPa, which is considerably lower than the structural steel fluence stress value (250 MPa). Therefore, the structure will resist the applied loads during its use (Figure 3).

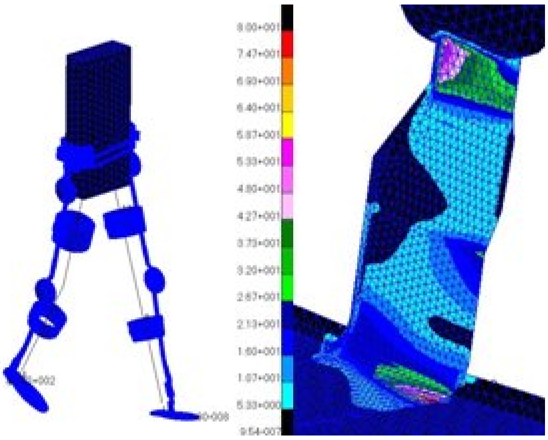


Figure 3: Results of the FE analyses. The maximum stresses are located near the ankle joint and the values are about 60 MPa.

4. Control design

The aim of the control design of the exoskeleton is the rehabilitation of patients through re-learning of neuromuscular movement patterns using robot aided training. As most of the patients using this exoskeleton are not able to manoeuvre the most common daily live movements by themselves, nor the rehabilitation movements, the control system has to drive the motors in a user transparent way through pre-calculated trajectories in time. This exoskeleton should be understood more as a movement assistant system than power assistant.

The control can be made through intention detection of the patient with modified patter recognition logics [10] and [11] depending on the disorder of the patient. In case of severe impairment, the rehabilitation movements would be with no user intention feedback but always with the target of stimulating user's interactuation with the exoskeleton.

The EMG signal is not only used to complement the control strategy and intention detection but also for neurofeedback application to the user during rehabilitation.

The first of all movement to start with is the standing up motion. Therefore the first experiments and control strategies developed have been designed for the stand up motion control. In future work more movements like gait, stair up, stair down, slope up and slope down walking movements will be approached.

4.1 EMG Signals

User's myoelectrical signals are acquired for monitoring, force calculation and intention detection issues.

The force of knee flexion and extension has been estimated as explained in [12] with a 200 ms RMS window to see differences between impaired persons EMG and non impaired persons. Therefore, a very simple test has been performed. Sitting in a table the user makes the maximal knee extension possible and maximal knee flexion possible, with no external help, and any upper limb motion in order to isolate the movement muscular origins and avoid any force contribution to knee extension and through hip movement. During the test EMG signals have been obtained with 1 Khz from DelSys 2.1 Differential Signal Conditioning Electrodes [15] of the vastus lateralis and biceps femoris. Also the knee angle position has been monitored. The data acquisition system is a NI DaqPad 6015 [16].

Electromyografic (EMG) signal and knee joint angle in sagital plane were measured in 10 non impaired patients and 8 impaired patients with different pathologies. These patients were recorded in the Guttmann Institute and Neurorehabilitation Hospital in Badalona, and the non impaired patients in Fatronik in San Sebastian according to a protocol approved by the local ethic committees.

4.2 Knee-calf-foot model

In order to obtain the knee joint torque from knee joint angle measurement in sagital plane, a simple knee-calf-foot model with two body segments (calf and foot) as shown in Figure 4.

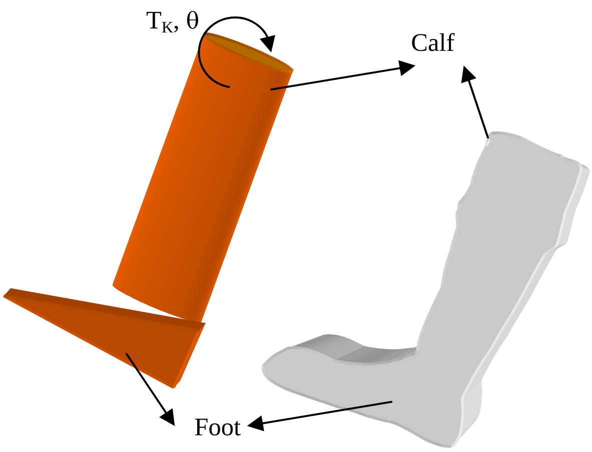


Figure 4.: Two body segment simulation

The corresponding inverse dynamic formula gives the necessary torque of the system responsible of recorded angle.

$$T_K = T_R + g\cdot\left(m_F\cdot L_F + m_C\cdot L_C\right)\cdot\sin(\theta) + \upsilon_K\dot{\theta} + \left(m_F\cdot L_F{}^2 + \frac{1}{3}m_C\cdot L_C{}^2\right)\ddot{\theta}$$

where m_F and L_F are mass of foot and length to the centre of gravity of the foot measured from the knee, m_C and L_C are mass of the calf and length to the centre of gravity of the calf measured from the knee centre and υ is the overall friction assumed 0.2 [Nms/rad][12].

4.2 Results

In Figure 5, the results of the EMG to Force calibration done by [12] is shown. The obtained force is reliable and with a total error of R=0.131 Nm/2.

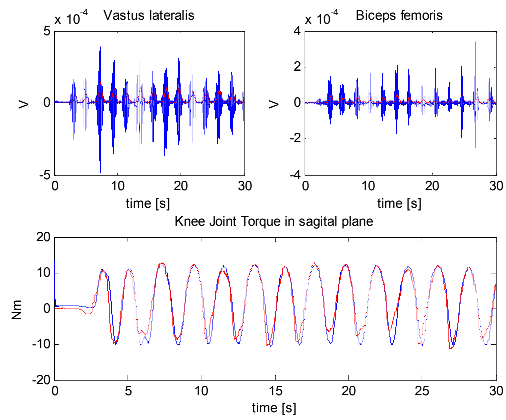


Figure 5: EMG-Force calibration of a 38 year old non impaired male during knee extension and flexion.

To see the effectiveness of this technique of EMG to force calibration, a research study was done with 8 patients with different type of spinal cord injuries. These patients were recorded in the Guttmann Institute and Neurorehabilitation Hospital in Badalona. All measurements have been done with explicit consent of the patients and according to a protocol approved by the local ethic committees.

The Figure 6 shows the result of the EMG to force conversion of a 31 year old partial sacral spinal cord injury patient. The result is an overall error of R=0.453 Nm/s.

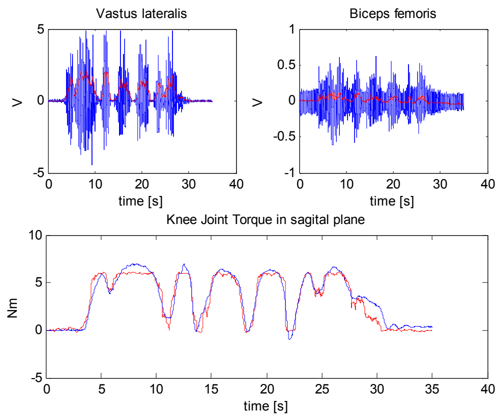


Figure 6: EMG-Force calibration of a 31 year old impaired male during knee extension and flexion. Partial sacral spinal cord injury.

4.3 Intention Detection

In order to do a position trajectory control of the knee joint during stand up motion the intention of standing up has to be detected before approaching the control of such movement.

The detection of the stand-up motion intention is based on the monitoring of the upper limb and hip angles.

At the beginning of the stand-up motion the upper body is bent forward to relocate the centre of gravity before activating the knee extension muscles. The point of maximal bent position (shown in Figure 7) is set as the beginning of the stand-up motion. Moreover, the duration of the centre of gravity relocation is proportional to the duration of the whole stand-up motion.

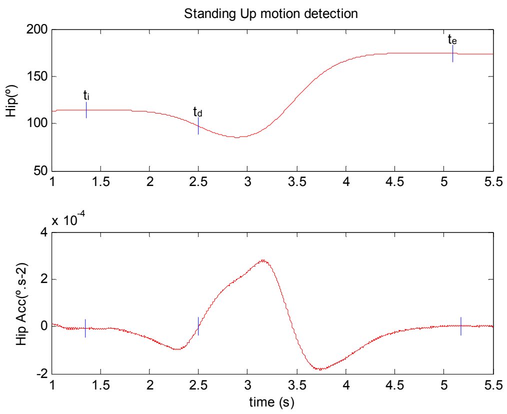


Figure 7: Intention detection of stand-up motion

The trajectory of the stand-up knee joint, has been modeled as a polynomial function of degree 7.

$$f(x)=\sum_{i=0}^{n} a_i x^i$$

The a_i constants of the polynom have been set experimentally using standing up knee joint trajectories of non impaired persons. These constants can also be adjusted recursively for each user.

5. Conclusions and further work

The exoskeleton is not targeting a specific pathology, but it will benefit patients with similar motor problems. Most of the potential users will be incapable to keep one's balance and therefore the circuits where the experimental tests will be made need to be closed tracks with balance keeping structural aids. Regarding this and using the developed mechanical prototype, a complete characterization of patients will be made, determining what characteristics will have the patients benefiting form the exoskeleton. This characterization will be carried out at Institute Guttmann.

Related to the detection of intention, it's impossible to define a sole actuation strategy, because the characteristics of the patients are very different. As a result, it's necessary to define some groups of users attending to the problems they have and the strategy of the detection of intention will be defined for each of these groups.

The exoskeleton has been designed to be used by patients, being able to give 50% of the required torque. This value was a necessary initial hypothesis to define the specifications of the drives, but after having built the exoskeleton, it will be validated experimentally. In any case, the criterion used to make the final torque selection assure having even a higher torque than this.

At the moment there are still some remaining issues in the development of the lower limb exoskeleton. In mechanical aspect, it's necessary to validate experimentally the design's comfort, the adaptability to the patient and the aesthetic effect that produces to the users.

On the other hand, the control design and the defined algorithms have to be extrapolated from one to four motors, the defined pattern trajectories have to be experimentally verified and the strategies of detection of the user's intention for the different groups of patients have to be defined.

6. Acknowledgements

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