

Exoskeleton design for functional rehabilitation in patients with neurological disorders and stroke

H. Zabaleta¹, M. Bureau¹, G. Eizmendi¹, E. Olaiz¹, J. Medina², M. Perez²

¹*Fatronik foundation, Research Technology Center. Donostia, San Sebastian, Gipuzkoa, Euskadi, Spain.*

²*Guttmann University Institute for Neurorehabilitation-UAB, Badalona, Spain*

¹{ hzabaleta, mbureau, geizmendi, eolaiz}@fatronik.com

²{jmedina, mperez.investigacio}@guttmann.com

Abstract—The prevalence of neurological disorders 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 functional 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 a whole mechatronic system has been done in order to approach functional rehabilitation in patients with neurological disorders and stroke. The EMG to force conversion in paraplegic patients is also described.

I. INTRODUCTION

THE prevalence of neurological diseases such as spinal cord injury, stroke, traumatic brain injury is increasing quickly in the industrialised societies. The improvements of immediate assistance protocols are increasing the survival rate in accidents and as a result, for example, the prevalence of the spinal cord injury has been doubled in the last 20 years. The prevalence and incidence per year of stroke, spinal cord injury and traumatic brain injury statistics in population of the United States are shown in the Table I.

TABLE I
PREVALENCE AND INCIDENCE OF NEUROLOGICAL DISEASES

Disease	Prevalence (per year)	Incidence (per year)
<i>Stroke</i>	1390	220
<i>Spinal cor injury</i>	900	205
<i>Traumatica brain injury</i>	200	175-200

Data obtained from American DHSS

Neuromuscular diseases cause mobility impairments and depending on the affection level, gait may be affected drastically making it even impossible. Rehabilitation process essential in progressing the recovery of people with diminished motor control skills. Adequate intensity and rehabilitation quality are key factors for successful rehabilitation.

The clinical characteristics severe physical impaired patient caused by neurological diseases, present big functional limitations. The rehabilitation should analyzes their needs in order to design a customized training and rehabilitation evolution. As each patient is a unique case of rehabilitation, a personal rehabilitation should be planed. Is essential to enhance the patients potentials with any technical aid system or orthopedic device, which will contribute to achieve a maximum level of functional independence.

The global objective of the work presented in this paper is the development of a portable 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 in upper limb extremities improve rehabilitation procedures, allowing setting up a more specific rehabilitation program, gaining motivation for the patient, independence from availability of occupational therapists, maximizing repeatability and precision [1][2]. In addition, game and robot mediated exercise regime increases motivation, practice volume and attention span during training [3]. It can also provide useful information to the rehabilitation staff about biomechanics and muscular status to determine the evolution of the patient.

In the last years several initiatives of robotic platforms for gait rehabilitation have been launched, but few have been successful and none of the ones targeting specifically lower limb functional rehabilitation is portable, reducing very much the possible applications.

Regarding this, Fatronik is developing a portable 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 of different pathologies, have balance, basic mobility and muscular

coordination problems.

In order to determine the specifications of the necessary exoskeleton drives, an initial maximal impairment estimation of the user was fixed to 50%. That is, the patient despite his or her impairment is capable of doing the 50% of the torque, that a non impaired person needs to perform in every daily live movements. This value will have to be reconsidered after experimental tests.

This paper presents the first year development of the described exoskeleton and research with medical staff of the Guttmann Institute and Neurorehabilitation Hospital in Badalona. 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. We include the conclusions of the developments and further work to be done.

II. STATE OF THE ART

There are various types of exoskeletons at research level. The most notable are the Hybrid Assistive Leg (HAL), developed by Prof. Sankai at the Cybernetics Laboratory of the University of Tsukuba in Japan [4] 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 which is also being used for people with walking disorders, but it's main target is not the functional rehabilitation of the lower limb. 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 was constructed 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 (hip 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 [5]. 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 based on sole pressure measurements, so there are no direct measurements from the user or human-machine contact areas [6], so the problems associated to interaction force or human muscle activity measurements disappear. This exoskeleton has 7 dofs driven with hydraulic cylinders: 3 at the hip, 1 at the knee and 3 at the ankle.

The RoboKnee is a powered knee brace that works parallel to the wearer's knee but does not transfer loads to the ground. This device transfers the load to the human

skeleton. The main characteristic of the RoboKnee is that the designed mechanical design and actuators achieve a high level of transparency. Through Series Elastic Actuators (SEA) very low impedance is reached. In this exoskeleton the user intent is determined through the ground reaction forces and joint angles [7].

The Lokomat and the LOPES are gait rehabilitation robots. The Lokomat is a four dof robotic orthosis [8] which proposes adaptive control methods that minimize the interaction forces with the patient with 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 [9]. Both systems designed to gait rehabilitation are non portable. The patient is attached to the system and doesn't move from the place, avoiding possible functional rehabilitation, which can be a key factor of the neuromuscular rehabilitation [10].

The state of the art at the moment is that there is no developed portable device for lower limb functional rehabilitation of patients with neurological disorders and stroke which could lead to new functional rehabilitation methods.

III. TECHNICAL DEVELOPMENT

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

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. This includes not only mechanical structure but also selection of drives. In order to have a preliminary exoskeleton to perform experimental analysis, a complete lower limb prototype has been designed.

The control system of the exoskeleton includes the method to detect user intention and decision making to assist patient in an adequate way during gait and stand up motion with the required torque in each joint. In order to achieve a robust system, the control system has been developed modularly [11][12]. 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.

IV. 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 at the same time, aesthetic issues requires compact and low weight drives.

A. Drive Design

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

The angle pattern data, forces and power of the lower limb joints in the sagittal plane have been largely studied [13][14]. The data of the gait cycle for the exoskeleton drive calculation have been obtained from different normalised gait experiments [13]. This 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), but it gives a enough precise base for a biomechanical estimation.

The values were computed considering a patient weight of 75 kg wearing a 25 kg weight exoskeleton. Joint moments in sagittal plane contribute to forward progression and dynamic balance, whereas, hip and ankle joint moments in frontal plane contribute to weight transfer from one leg to other and to lateral balance of head/arms/trunk (HAT) segment of body [15].

If the patient has balance control through well controlled HAT movements or with some external balance aid system, the lack of force in the sagittal plane will impede to perform most of the daily live movements. 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.

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

The volume, the weight and the aesthetic also have to be considered in the drive selection. The installation and tuning of the final solution should be easy and fast.

TABLE II
MAXIMUM POWER OF HIP, KNEE AND ANKLE

Joint	Movement	Power (W) Gait	Power (W) upstairs	Power (W) Stairs down	Max Power (W)
Hip	Flexion	82,5	87,75	33,75	87,75
	Extension	-42,3	0	-114,7	
	Abduction	26,2			26,2
	Adduction	-25,5			
	Ext. Rotation	4,9			
Int. Rotation	-2,2			4,93	
Knee	Flexion	67,4	195,8	13,5	195,75
	Extension	-131	-20,2	-325	
Ankle	Flexion	165	195,75	67,5	195,75
	Extension	-39,5	-13,5	-249	
	Abduction	2,97			2,97
	Adduction	-0,81			

Data obtained from Winter *et al* [19].

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. Once all 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.

The second 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, that 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 choice are the electrical motors. This choice 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 and mechanically.

In this kind of actuation, the high torque requirements of the joints 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 reaching the objective values has been designed. The estimated output torque is 71 Nm, in a 5.14 rad/s speed and with a total power of 450 W. The lower power Robodrive motor kit would be limited in speed.

Finally the designed drive shown in Fig. 1 has a diameter of 113 mm, a 104.5 mm length at weighs 2.8 kg. Another factor to choose electrical motors is that the background on them is much higher.

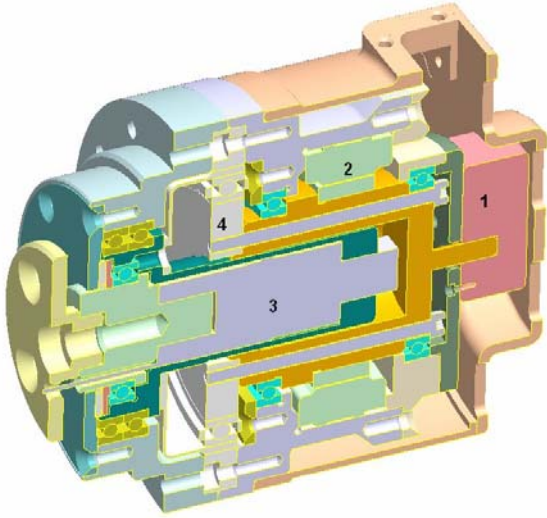


Fig. 1: Main parts of the final solution. 1: Encoder; 2: Rotor and Stator
3: Torque sensor and 4: Harmonic Drive compact reductor

B. 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 [16]. 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 is 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 were obtained from K. Luttgens, and N. Hamilton [17] (Table III).

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.

TABLE III
ACTUATED DEGREES OF FREEDOM

Joint	Angle (°)	Torque (Nm)	Power (W)	Velocity (rpm)
Hip	-10 / 75	72	87,75	25,78
Knee	3,2 / 100	101,25	195,75	45,83

It's especially important to assure that the knee joint position doesn't exceed an angle of 0°. Apart from the mechanical stops 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 orthopedics specialist, using materials and shapes that assure the patient comfort but rigid enough to guarantee the transmission of the movement to the patient.

V. CONTROL DESIGN

The aims of the control system of the exoskeleton are several.

First and most important is to achieve a control model in order to drive the structure in such a way that maximizes 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 an 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 pattern recognition logics [18] and [19] 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 users interaction with the exoskeleton.

The EMG signal can also be used to complement the control strategy and intention detection.

On the other hand the control and data acquisition and processing unit can also be used for neurofeedback application to the user during rehabilitation, visual or acoustic feedback and patients muscular activity monitorization.

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.

A. EMG Signal based force control

As in other exoskeletons [5]-[7], users myoelectrical signal is acquired for monitoring, force calculation and intention detection issues.

As the final users of this exoskeleton will have high level of muscular atrophy the EMG to force conversion algorithms used in non impaired cases have to be validated first. The force of knee flexion and extension has been

estimated as explained in [20] with a 200 ms RMS window to see differences between impaired persons and non impaired persons electromyographic (EMG) signal. 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 muscular origins of the movement 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 [21] of the vastus lateralis and biceps femoris. Also the knee angle position has been monitored. The data acquisition system is a NI DaqPad 6015 [22].

EMG signal and knee joint angle in sagittal 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.

B. Knee-Calf Model

In order to obtain the knee joint torque from knee joint angle measurement in sagittal plane, a simple knee-calf-foot model with two body segments (calf and foot). For this novel approach, the system was simulated as a point mass (foot) attached to a slender rod (calf).

The corresponding inverse dynamic formula (1) gives the necessary torque of the system responsible of recorded angle.

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

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 of the calf measured from the knee centre and v is the overall friction assumed 0.2 [Nms/rad][20].

C. 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 as shown in Fig. 2. All ankle, knee, and hip angles are monitored with DelSys S700 goniometers, as well as the evolution of a representation the centre of gravity based on 4 flexiforce [23] A201 4.4N pressure sensors positioned in the foot sole. Each one is located under the heel, the head of the fifth metatarsal, the head of the first metatarsal, and the great toe [14]. The data are stored at 1KHz with a NI DaqPad 6015

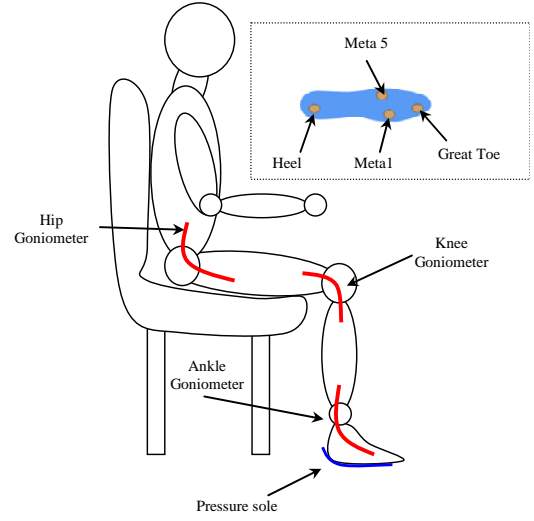


Fig. 2: Experiment setup for intention detection of stand-up motion [22].

At the beginning of the stand-up motion the upper body the upper limb is bent forward to relocate the centre of gravity before activating the knee extension muscles. The point of maximal bent position (t_3 in Fig. 3) is set as the beginning of the stand-up motion. This point is characterized by a hip acceleration change (i.e. acceleration equal to zero).

Moreover, the duration of the centre of gravity relocation (t_1-t_0) was experimentally found proportional to the duration of the whole stand-up motion (t_4-t_0). The pattern activation time (t_4-t_3) can be deduced using this proportionality.

The trajectory of the stand-up knee joint has been

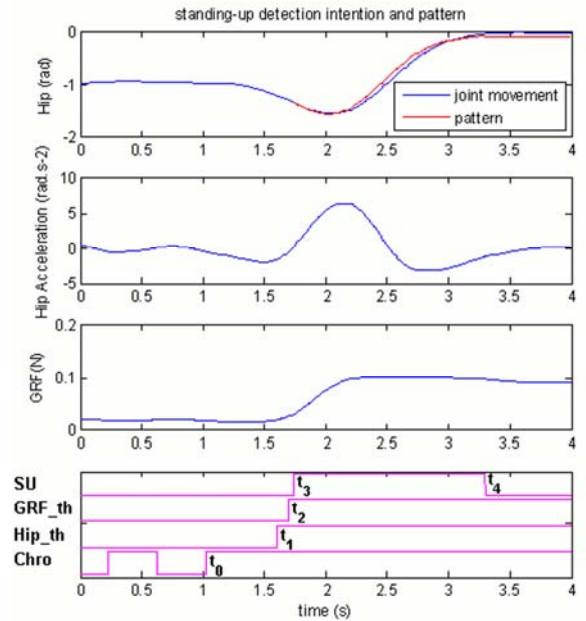


Fig. 3: Intention detection of stand-up motion. Chro: chronometer initialized at each potential start of standing-up motion. Hip_th: hip angle threshold verification. GRF_th: GRF threshold verification. SU: Stand-up pattern activation.

modelled as a polynomial function of degree 7.

$$f(x) = \sum_{i=0}^n a_i x^i \quad (2)$$

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

VI. CONCLUSIONS AND FURTHER WORK

The exoskeleton is not targeting a specific pathology, but it will benefit patients with similar motor problems. Regarding the patients, a biomechanical analysis has been made in order to determine what characteristics common will have the patients benefiting from the exoskeleton (weakness, balance control, upper limb mobility control and proprioception). This characterization has been carried out at Institute Guttmann.

A. Validation of EMG-force based control

The results of the EMG to Force calibration done by [20] in non impaired persons show to be very reliable. The obtained force has a total error of $R=0.171 \pm 0.092$ Nm/s.

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 Fig. 4 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 $R=0.521 \pm 0.155$ Nm/s.

As shown in Fig. 4, the EMG to force calibration is clearly extrapolable to severe mobility impaired patients after a partial spinal cord injury. Although the recorded signals have much lower amplitude, the correlation between

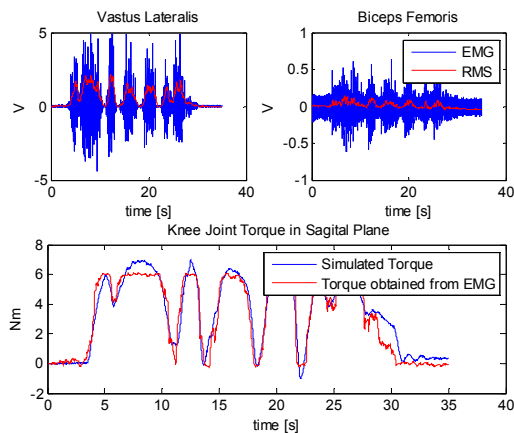


Fig. 4: EMG-Force calibration of a 31 year old impaired male during knee extension and flexion. Partial sacral spinal cord injury. $R=0.521 \pm 0.155$ Nm/s.

force and EMG signal is still present.

On the other hand as many physically disabled suffer from neuronal damage (e.g., stroke or spinal cord injury) that prohibits or disturbs the control of movements or muscular force control [14], a force feedback [4][5][6][7][24] and [25], control is unlikely to be done in such patients.

Other control methods such as rule based control, or intention detection based control could be used for such patients.

B. Validation of Stand Up detection

The intention detection of stand up motion has been tested in two severe impaired users (wheelchair users) in different conditions (standing up without external aids, walker aided standing up motion, i.e.). These test were performed in two patients (Table IV) in the Guttmann Institute and Neurorehabilitation Hospital, according to a protocol approved by the local ethic committees.

The standing up trials were performed 10 times in different conditions:

-- Without any external help

TABLE IV
PATIENT CHARACTERISTICS

Patient	Sex	Age	Weight	Height	Injury
Nr. 1	Male	37	70	181	T7 (Grade A)
Nr. 2	Female	49	72	175	T7 (Grade C ~ D)

Age in years; Weight in Kg; Height in cm; Injury according to ASIA / IMSOP

-- With partial weight support on armrests

Patient Nr. 1 was asked only to try to perform this trials, as he wasn't able to perform them completely. Patient Nr. 2 was able to perform the tasks completely only in the second case.

In case of the first patient the extreme movement impairment avoided standing up detection. On the other hand, in the case of the second patient, each of the trial was detected successfully without any previous training. No further attempt was done to see how difficult it is to train the motion in such way that the task could be detected by the algorithm, but further development is being done to achieve the standing up motion using the recorded signals "off line".

In case of severe impaired (wheelchair users) persons no successful sole actuation strategy to detect standing up motion could be achieved.

C. Exoskeleton design conclusions

The exoskeleton has been designed to be used by patients with a maximum lower limb disability of 50%. 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.

The EMG-force based control design as explained before is unlikely to be successful in severe impaired patients. Another possible control mode could be a EMG triggered

motion control, but it hasn't been programmed nor tested yet. The defined algorithms of intention based control 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.

Most of the potential users are 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. This self balance control is a key factor for functional rehabilitation.

All exoskeleton devices we have information about are designed either for persons who have good balance control or have some walker or other extra device with aids the patient on his balance control [4]-[9] and [26]-[28]. For example, a motored walker which would support partially the patient's weight and enhance his balance skills. This walker would reduce significantly the needed drive power and would make the exoskeleton lighter.

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