2017 International Conference on Rehabilitation Robotics (ICORR)
QEII Centre, London, UK, July 17-20, 2017.

An Assistive Lower Limb Exoskeleton for People with Neurological Gait Disorders

A. Ortlieb, M. Bouri, IEEE senior member, R. Baud, and H. Bleuler, IEEE member

Abstract— **Lower limb exoskeletons have already proven the capability to give back mobility to people suffering from spinal cord injury (SCI). Other important populations such as people with multiple sclerosis or muscular dystrophy, frail elderly and stroke victims, suffer from severe gait impairments and could benefit from similar technology. The work presented in the current paper describes a novel design of a 6-actuated degrees of freedom (DOFs) assistive lower limb exoskeleton for people with moderate mobility impairments. The electrical actuators are all remotely located on the back of the user for a more compact design with high dynamics. Cable driven solutions are used to transmit the flexion/extension of the hip and knee joints, while a powerful ballscrew carries out the hip adduction/abduction. The design of this exoskeleton, named AUTONOMYO, follows the key specifications of being highly back-drivable and able to perform dynamic motions at low energy consumption. AUTONOMYO is capable to assist the user's balance by providing complementary torques at the hip and the knee. Results show that the projected level of assistance for sit-to-stand transition varies from 50% to 100% in function of the user's bodyweight and height while higher level of assistance are reached for walking and stairs climbing activities.**

I. INTRODUCTION

Locomotion disorders can severely affect the ambulatory capacity of individuals and result in serious issues for the persons concerned and for the society. Based on the *2012 U.S. Health National Survey* [1], 7% of the population sample reported not to be able (or to find "very difficult") to walk one quarter of a mile (~400m). Serious mobility difficulties thus concern a large part of the population (estimated about 22 million people in the USA). With a median about 64 years old [1], gait disorders largely affect elderly people and is subject to a continuous increase because of the population ageing. Regarding the other half of the affected population which is of working age, the unemployment rate is reported to be over 75% [1] and is a significant burden for the society.

Among the different causes of walking impairments, most of them are related to neurological disorders such as:

- Stroke (incidence: 11–35/10'000 /year [2])
- Parkinson disease (PD) (prevalence: 10–20/10'000 [3])
- Multiple sclerosis (MS) (prevalence: ~1-15/10'000 [4])
- Spinal cord injury (SCI) (prevalence: 2.5–9/10'000 [5])
- Neuromuscular diseases (NMD) (prev.: ~3/10'000 [6])

*Research supported by ASRIMM (Association Suisse Romande Intervenant contre les Maladies neuro-Musculaires) and FSRMM (Fondation Suisse de Recherche sur les Maladies Musculaires).

A. Ortlieb (amalric.ortlieb@epfl.ch), R. Baud (romain.baud@epfl.ch), M. Bouri (mohamed.bouri@epfl.ch) and H. Bleuler (hannes.bleuler@epfl.ch) are with the Laboratory of Robotic Systems from Ecole Polytechnique Fédérale de Lausanne, Switzerland.

Besides strokes for which physical rehabilitation allows to recover capabilities, most neurological disorders are currently incurable. PD, MS and NMD patients lose their motor and functional skills slowly while it is abrupt for SCI and stroke patients. Moderate to severe symptoms lead, among other, to a loss of ambulation due to poor balance control and weak muscles and to the loss of autonomy. Wheelchairs are the unique alternative for such patients. However, recent technological advances in wearable robotics have opened the door to a new era of walking aids: medical lower limb exoskeletons.

Lower limb exoskeletons have first been developed for various applications such as strength augmentation, load carrying and vertical mobilization of paraplegics. Vukobratovic et al. had already developed a design of exoskeleton for SCI patients in the 70's [7]. However, the first certified products have been only available on the market since 2010 (ReWalk Robotics: CE marked in 2010 and FDA cleared in 2014, Ekso bionics: FDA cleared in 2012 and CE marked in 2012 and 2016(medical), Cyberdyne HAL: CE marked in 2013(medical) [8]–[10]). While the number of proposed medical devices is in constant expansion, the diversity of targeted populations is confined mostly to people with severe and partial impairments: i.e. SCI patients and victims of stroke.

The current paper is presenting a novel design of a lower limb exoskeleton (Fig.1) that targets people with moderate neurological disorders [11]. This device is intended to be suited to neurological disorders such as PD, MS, NMD or stroke. Original design specifications are expressed in the following chapter where the differences and implications of mobilization and assistive strategies are described. The design and characteristics of the AUTONOMYO exoskeleton are

Figure 1. AUTONOMYO exoskeleton worn by a healthy subject, the current prototype is missing leg attachments (see Fig.2)

depicted in part three before assessing the capabilities and limitations of the electro-mechanical elements. Tests include the evaluation of resistance under maximal loads and the energy consumption during gait trajectories at different walking speeds.

II. Design Specifications

A. Assistance vs. Mobilization Strategies

Mobilization refers to a strategy where a device is controlled in position to follow some predefined trajectories; each corresponds to a predefined gait cycle. A mobilization strategy considers only unidirectional physical interactions from the exoskeleton to the user's legs. It typically fits with the use by SCI people who are not capable to, fully or partially, control over their lower limbs. That strategy is suited to work with crutches thanks to which the user is able to balance and orientate the direction of walking. One drawback, however, is the need of non-impaired upper limbs for a good use of crutches.

In opposition, assistance refers to a strategy where the user guides the motion while the device provides additional forces to perform the motion. In this case, bidirectional interactions between the user and the exoskeleton have to be managed. The main advantage of the assistive approach is that it encourages the implication of the user when walking with the exoskeleton. This latter could aid the patient perform a task while offering rehabilitation benefits. If assistance allows the user to fully manage in real time the motion she/he is performing, the controller is however sensitive to the variability in the users' level of impairment. Considerable work still need to be done in that sense.

Considering the mechanical specifications, a strategy of assistance requires a high transparency (low impedance of the mechanism) which implies:

1. Good backdrivability
2. Low inertia perception

These specifications concern mainly the actuators and transmission mechanisms of the exoskeleton.

B. Kinematics and Dynamics

Regarding the panel of activities aimed to be performed, i.e. level walking, stairs climbing/descending and sit-to-stand transition; kinematic and dynamic specifications can be defined from the literature. The range of motion for each joint should respect the overall maximal and minimal values recorded by one of the activities. The higher flexion angles for the hip and the knee are obtained during the sit-to-stand transition, where angles' peaks reach 104° and 106° respectively for the hip and the knee flexion [12], [13]. Maximal extension angles for the hip are obtained during fast walking at toe off with peak about -20° of flexion [14], [15]. Knee extension maximal angle is physiological and is reported about -1° to -10° of flexion [16]. Hip adduction/abduction angle ranges about ±5° [17].

In terms of dynamics, two criteria need to be met: the higher gait cycle rate and a defined percentage of assistance. These constraints determine specifications of peak velocity, cyclic acceleration cost, nominal torque and peak torque. As presented [18], the limit of walking capacity in NMD patients is situated about a need of 50% of assistance to recover a healthy walking ability. Table I presents quantitative values regarding walking extracted from measurements of Ounpuu 1994 in healthy gait kinematics and kinetics at speed of 1.17 m/s (4.2 km/h) and mean cycle duration of 0.9 s [17], data for the hip adduction/ abduction are taken from Schache and Baker 2006 [19]. Table II and III respectively illustrates values for the sit-to-stand transition and the ascension of stairs based on the studies from Mak et al. 2003 and Protopapadaki et al. 2007. Note that loads are normalized over bodyweight and also height when rising from a chair as large angles (about 90°) are involved. The requirements of the load capacity of the actuators not only depend on the level of force provided to the user (the assistance), but also on the force consumed to perform the motion of the device. As reported in [20], dynamics of walking can require a full motor capacity without load.

TABLE I. Dynamic Specifications for Level Walking Based on [17] and [19]

Joints	Specifications in Level Walking		
	Peak Velocity [°/s]	***50% of RMS Torque [Nm/kg]^d***	***50% of Peak Torque [Nm/kg]^d***
Hip A-A^a	55.7	0.18 (9/18 Nm)^e	0.33 (16.5/33 Nm)^e
Hip F-E^b	178.5	0.15 (7.5/15 Nm)^e	0.36 (18/36 Nm)^e
Knee F-E^b	360	0.09 (4.5/9 Nm)^e	0.25 (12.5/25 Nm)^e
Ankle DF-PF^c	201.6	0.25 (12/25 Nm)^e	0.61 (30.5/61 Nm)^e

a. A-A: adduction/abduction, b. F-E: flexion/extension, c. DF-PF: dorsiflexion/plantar flexion
d. Torques are normalized over bodyweight, e. Torque values for a bodyweight of 50kg/100kg respectively

TABLE II. Dynamic Specifications for Rising from a Chair Based on [13]

Joints	Specifications in Sit-to-Stand (STS) Transition		
	STS duration [s]	***50% of Peak Torque per Leg[Nm/kg/m]^d***	***Total Peak Torque per Leg [Nm/kg/m]^d***
Hip A-A^a	1.4 in healthy persons (2.3 in persons with Parkinson)	-	-
Hip F-E^b		0.23 (18/43 Nm)^e	0.45 (35/87 Nm)^e
Knee F-E^b		0.29 (23/56 Nm)^e	0.58 (45/111 Nm)^e
Ankle DF-PF^c		0.16 (12/30 Nm)^e	0.32 (25/61 Nm)^e

a. A-A: adduction/abduction, b. F-E: flexion/extension, c. DF-PF: dorsiflexion/plantar flexion
d. Torques are normalized over bodyweight and body height and assumed symmetrical in both legs, e. Torque values for a bodyweight and height of 50kg and 1m55/100kg and 1m90 respectively

TABLE III. Dynamic Specifications for Ascending Stairs Based on [21]

Joints	Specifications in Ascending Stairs		
	Cycle duration [s]	***50% of RMS Torque [Nm/kg]^d***	***50% of Peak Torque [Nm/kg]^d***
Hip A-A^a	1.45	-	-
Hip F-E^b		0.17 (8.5/17 Nm)^e	0.38 (19/38 Nm)^e
Knee F-E^b		0.14 (7/14 Nm)^e	0.29 (15/29 Nm)^e
Ankle DF-PF^c		0.38 (19/38 Nm)^e	0.72 (36/72 Nm)^e

a. A-A: adduction/abduction, b. F-E: flexion/extension, c. DF-PF: dorsiflexion/plantar flexion
d. Torques are normalized over bodyweight, e. Torque values for a bodyweight of 50kg/100kg respectively

C. Ergonomy

Neurological disorders have the specificity of not being restricted to the lower limbs of the body but to a large part of it, except for SCI patients with lowest lesions. Hence, the use of external support such as crutches is not adapted for such people and balance should be carried out by other means. Ergonomic aspects such as ease of don and doff, autonomy of several hours, weight and compactness of the device for transportation and handling are to be considered. Another aspect that is not treated in this paper is the sound level of the actuation that has to be as low as possible. Eventually, safety is also a key aspect to be considered in the design of an exoskeleton as presented in [22].

III. Design of Autonomyo

A. Global Architecture

The exoskeleton presented in this paper, named AUTONOMYO, is composed of three actuated degrees of freedom (DOFs) per leg, corresponding to the human hip adduction/abduction, hip flexion/extension and knee flexion/extension joints. Three passive DOFs per leg are located about the ankle and reproduce a ball joint with a variable stiffness and viscosity on each axis. The exoskeleton is fastened to the wearer through a maximum of three physical interfaces per leg at the foot, shank and thigh, plus one interface at the trunk level (Fig. 2). The adequate position and number of physical interfaces are not discussed in this paper and will be subject to further investigations.

The electronics includes three motor boards (drives) specifically designed at the laboratory of Robotic Systems (EPFL, Lausanne) for the low level control of the six actuators and a Beagle Bone Black (BeagleBoard.org®) CPU board manages the hig level controller (more detail can be found in [23]). The device is empowered by a source of 16 Ah Lithium polymer batteries at 48V for a total weight of 3.8 kg. The total weight of the exoskeleton is about 22.5 kg, including the batteries. About 2/3 of this weight (15 kg) is located in the upper part of the exoskeleton, between the hip and the chest of the user.

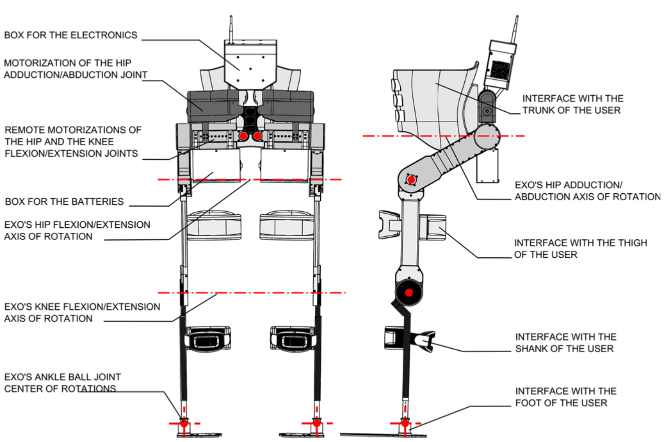

Figure 2. Back and side view of the 6 actuated DOFs exoskeleton's design

B. Hip Add/Abduction Actuation

The actuation of the hip adduction/abduction is part of the originality of the present design as only few devices are equipped with this actuated DOF. The angular range of motion during walking is small, about 10° but the torques required are similar to the one of the hip flexion/extension during walking. Despite a small range of motion, the hip adduction/abduction interests consist in its partial implication in the lateral balance control [24], plus the high correlation between its strength and gait velocity [25] in the human body.

The hip adduction/abduction mechanism is based on a planar four-bar linkage made of three rotations and one translation as illustrated on Fig. 3. The torque from the motor and gear unit is transmitted through a belt and pulleys system to a ballscrew that actuates the four-bar linkage. Eventually, it results in the rotation of the hip adduction/abduction joint.

A brushless DC motor (EC-4pole 30, Maxon Motor AG, Switzerland) is selected for its high power density and small inertia. It is combined with a planetary gear (GP32 HP, Maxon Motor AG, Switzerland) of transmission ratio i = 14:1 which allows to reduce the perceived inertia of the ballscrew at the motor of a factor i^2 = 196. The mounted ballscrew (FA compact series, NSK Ltd, Japan) has a pitch of 5mm for a stroke of 50mm. The nut is fastened to a linear guide (prismatic joint on Fig.3) to avoid radial loading of the ballscrew. An encoder with 1024 pulses per turn measures the motor position, while a linear potentiometer positioned around the final joint provides the absolute hip position. The range of motion for the hip joint is +15° in adduction and +25° in abduction. The total transmission ratio from the motor to the hip is quasi linear within the range of motion (maximal deviation of 4.5%) and equals to i = 1'592:1. Hence, the nominal torque at the hip level, without any energy losses at the motor and transmission level would be of 148 Nm with the currently mounted motor (motor nominal torque: 92.9 mNm). With a nominal motor speed of 16'600 rpm at a volage of 48V, velocities of up to 625°/s are reached at the joint, which is largely over the specifications stated in Table I. The aspects of transparency (backdrivability of the mechanism), dynamics and fulfillment of specifications for given tasks are evaluated in the next chapter.

C. Hip and Knee Flex/Extension Actuation

The actuation units of the hip and knee flexion/extension are similarly designed. Each unit consists of one brushless motor

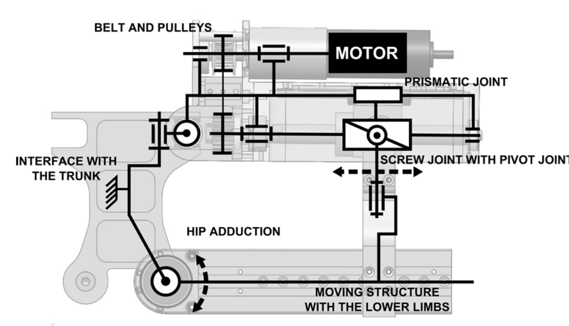

Figure 3. Kinematic scheme of the hip adduction/abduction mechanism

(EC-i 40, Maxon Motor AG, Switzerland) and a corresponding gearbox (GP42 HP, Maxon Motor AG, Switzerland) with a 66:1 transmission ratio. The actuation units are remotely located about the trunk of the user and the power is transmitted to the joints using a cable-pulleys mechanism as illustrated on Fig.4. A mechanical reduction of 3:1 is additionally provided at this stage thanks to the ratio of diameters between pulleys. A very flexible stainless steel wire-rope of diameter 1.76mm, with minimum breaking load of 2100 N (Carlstahl Technocables, Germany) is used. As presented on Fig.4, the hip and knee flexion/extension are driven over two stages: i) from the motors to the hip joint and ii) from the hip joint to the knee joint.

This design introduces a kinematic coupling between the hip and the knee joint where a positive ratio of 1:1 between both flexions has been determined as optimized in regard of the kinematics and control of the three activities of level walking, sit-to-stand transition and stairs climbing. Hence, a hip flexion motion, driven by motor B (Fig. 4), results in an equal and simultaneous knee flexion motion, whereas motor A drives only the knee flexion. The forward and inverse kinematics models are given in equations (1) and (2) respectively.

$$\begin{pmatrix}\dot{\theta}_{HIP}\\ \dot{\theta}_{KNEE}\end{pmatrix} = J\cdot\begin{pmatrix}\dot{q}_B\\ \dot{q}_A\end{pmatrix}\cdot\frac{1}{i},\ with\, J = \begin{pmatrix}1 & 0\\ 1 & 1\end{pmatrix} \tag{1}$$

$$\begin{pmatrix}\Gamma_{HIP}\\ \Gamma_{KNEE}\end{pmatrix} = J^{-1}\cdot\begin{pmatrix}\Gamma_B\\ \Gamma_A\end{pmatrix}\cdot i,\ with\, J^{-1} = \begin{pmatrix}1 & 0\\ -1 & 1\end{pmatrix} \tag{2}$$

With $\dot{\theta}$ the joints velocities, $\dot{q}$ the motors velocities, Γ the torques at the motors and joints and i the transmission ratio. The range of motion by design for this exoskeleton is -30° to 120° for the hip flexion and -10° to 100° for the knee flexion. The total transmission ratio between motors and joints is i=198:1 which allows to reach a nominal torque of ±40 Nm for the hip flexion and ±40 Nm – Γ_{HIP} for the knee flexion. Peak torques are limited by the minimal breaking load of the cables that theoretically correspond to an amplitude of 84 Nm per joint. Considering the preloading of the cables, the peak torques are estimated about 60 Nm. The selected motors are limited to a velocity of 8000 rpm, which corresponds to a maximal velocity of ±242 °/s at the hip and ±242 °/s + V_{HIP} at the knee. Eventually the effect of coupling between the hip and knee joints affects the forces transmitted to the ground or to the user.

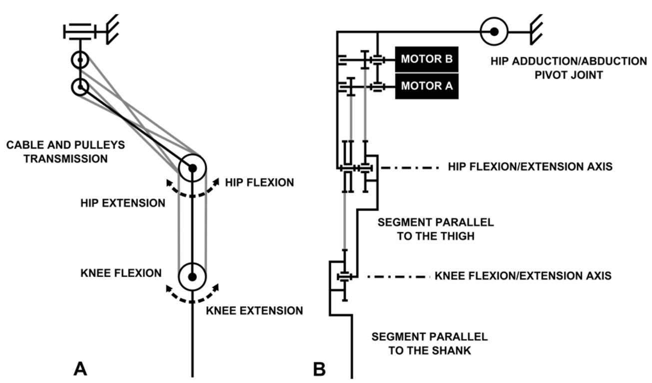

Figure 4. Kinematic scheme of both hip and knee flexion/extension mechanisms. A. View from the side (sagittal plane). B. View from front.

In Fig.5, the spans of forces applied by the exoskeleton at motor nominal torque and for different knee flexion angles are illustrated. The three cases of: no coupling, positive coupling (as described by (1) and (2)) and negative coupling (where $\Gamma_{KNEE} = i\cdot\Gamma_A + i\cdot\Gamma_B$) are compared. It shows that the horizontal span of forces increases with the knee flexion but is not sensitive to the coupling. However, the vertical span of forces largely increases close to the knee full extension (joint singularity) and is 166% larger for the negative coupling compared to the two others. The span of forces without coupling and with positive coupling are almost identical but one pushes stronger forward and the other backward relatively to the hip joint. The negative coupling design seems promising regarding its broader span of forces, however it implies high velocities at synchronous hip and knee flexion.

Remotely locating the motor units near the trunk of the user allows diminishing the thickness of the segments along the legs down to 24 mm. It also permitted to reduce the inertia at the hip joint of about 30% and offer a better transparency through the last stage of reduction made by cables.

IV. Evaluation of the Actuation Mechanisms

The evaluation is to define if the exoskeleton fulfill the requirements regarding the three main activities of level walking, stairs climbing and sit-to-stand transition. The limitations are peak velocities, peak torques and total RMS torques. Transparency of the system is also partially addressed to see how much energy is required by the user to move as the exoskeleton would not be powered.

A. Method

Both assessments described below are performed on one side (one leg) of the exoskeleton while it is rigidly fastened to a stiff structure at the trunk level (trunk, thigh and shank interfaces are dismantled). The motion is thus performed at the foot level and the exoskeleton is tested without being worn by somebody.

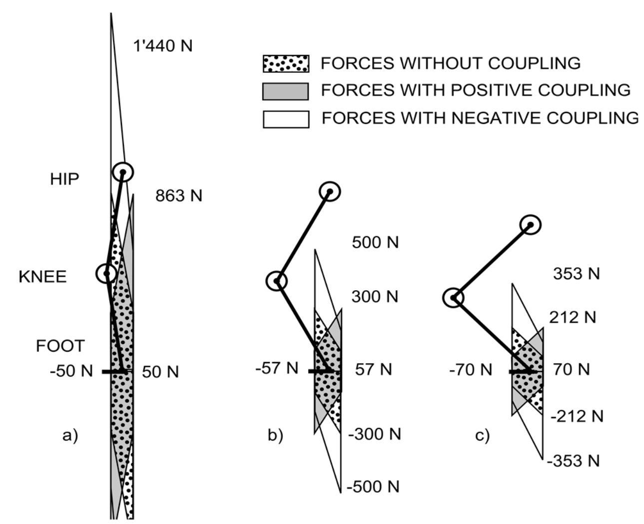

Figure 5. Range of forces at nominal use (40 Nm) for different coupling scenarios and knee flexion angles. a) Range with 20° of knee flexion, b) with 60° of knee flexion and c) with 90° of knee flexion.

1) Evaluation of peak torque

In order to validate the resistance of the mechanism to peak torques of 60 Nm at the hip and knee flexion/extension joints, the controller is set to position mode and assigned to different static positions. Then a force, corresponding to the peak torque is exerted using a dynamometer at the foot level. Pulling forces are exerted repetitively by hand 10 times during at least 30 s each time.

2) Evaluation of dynamical behavior

The dynamical behavior allows to identify the impedance of the system which provides; i) the amount of power consumed by the motors that is not transmitted to the user (e.g. viscous friction or inertia at the motor) and ii) the transparency of the system. To measure the dynamical behavior; the exoskeleton without additional load is controlled in position with a cyclical gait trajectory pattern (gait pattern from Ounpuu 1994 [17]) and all joints are actuated together. Trials are performed for different gait cadences; i.e. 52, 76, 93 and 108 steps/min, corresponding respectively to walking velocities of; 1km/h, 2km/h, 3km/h and 4km/h based on Stoquart et al. [26].

Kinematics are extracted and filtered from the encoders located on the motors and are eventually converted in the joint coordinate to get each joint position, velocity and accelerations. Motor torques are obtained from the target current given by the controller and the current is also measured directly on the motorboards. Lithium-polymer batteries with 16Ah at 48V are used to power the motherboard and the motorboards. Because of the motorboards design, the system is limited to a maximal tension of 50V.

B. Results

Peak torques of 60 Nm have been repetitively measured on the hip and knee flexion/extension joints of the exoskeleton. No sign of failure have been detected at such torques. The hip adduction/abduction joint maximal torque has not been experimentally validated since it is difficult to reproduce a load of 148 Nm and the elements involved in the transmission are not solicited close to their load limit.

One single gait trajectory but at different frequencies has been reproduced by the exoskeleton fastened to a table to measure the RMS torques that are consumed by the actuated joints before being able to provide any assistance. Results for walking cadences between 50 and 110 steps per minute are illustrated on Fig.6 for each motor of the three joints. The gait trajectories are reproduced with a mean error below 2° but at the knee flexion/extension joint. Due to the motorboard limitation to 50V, the knee full extension preceding the heel strike during walking reaches the maximal velocity already at a cadence of 70 steps per minute (slow walking). Values for the knee RMS torque presented on Fig.6 are simulated based on the characterized impedance of the exoskeleton's knee joint.

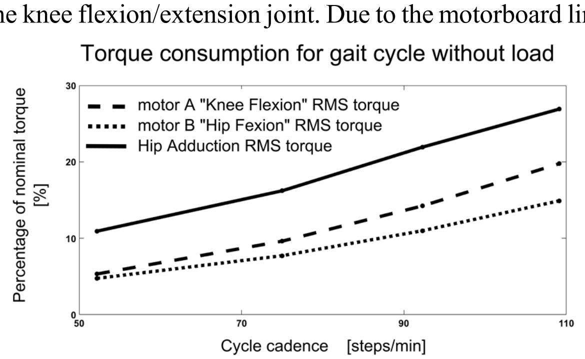

Figure 6. Percentage of motors nominal torque needed to perform gait cycles at different cadences

The hip adduction/abduction consumes a higher RMS torque relatively to its nominal capacity compared to the other joints: from 10% to 25% between slow and fast walking. This can be explained by the high reduction ratio that greatly increases the accelerations at the motor level. Hip and knee flexion/extension RMS torques range from 5% to 20% of the nominal capacity for slow to fast walking velocities.

C. Capacity of Assistance

Based on the experimental results, the remaining torque capacity at each joint can be calculated to evaluate the maximal level of assistance that can be provided. The limiting factors are the nominal torques and peak torques supported by each actuated joint compared to the requested torques during the different activities.

1) Level Walking

Considering as reference a comfortable cadence of 92 steps per minute during walking, the remaining nominal torques at each joint are respectively of {115, 36, 34} Nm for the hip adduction/abduction and the hip and knee flexion/extension. Based on level walking torques presented in Table I, an assistance of 100% can be provided for people up to 85kg. Over that bodyweight, the maximum level of assistance decreases linearly.

2) Sit-To-Stand Transition

Referring to torque specifications presented in Table II, the exoskeleton can provide an assistance of 100% for a person of 1m55 and 50kg while the level of assistance decreases inversely with body height and bodyweight to reach about 50% for a person of 1m90 and 100kg.

3) Stairs Climbing

Following specifications given in Table III, the level of assistance during stairs climbing is limited by peak torques about the hip flexion/extension. An assistance of 100% can be provided up to a bodyweight of 75kg, over that bodyweight, the affordable level of assistance decreases linearly.

V. Discussion

The aim of the current design is to propose a lower limb exoskeleton device targeting people with moderate gait impairments. Candidates are people affected by neurological disorders such as people with muscular dystrophy, multiple sclerosis, Parkinson's disease or stroke after-effects. The device need to be highly back-drivable with a large freedom of motion not to constrain the movements of the user. In addition, equipped actuators should be able to provide the adequate power to enhance in a natural manner the daily mobility activities: level walking, sitting and standing, ascending and descending stairs. Tests on the torque requirements to perform gait trajectories in no-load condition reveal a good backdrivability for the hip and knee flexion/extension joints but a high impedance at the hip adduction/abduction joint. At a comfortable cadence (92 steps per minute) between 11% and

23% of the nominal torque capacity of each actuation unit are consumed only to perform the motion of the exoskeleton. Accordingly, an important part of the torque remains and can be assigned to assistance. The sit-to-stand transition is the most demanding activity in terms of peak torques compared to both level walking and stairs climbing. The level of assistance that can be provided highly depends on the height and bodyweight of the user. The current design can assist between 50% to 100% depending on the user parameters which coincide with the need of most of the targeted users.

VI. CONCLUSION

The current paper exposes the joints' requirements to design a lower limb assisting exoskeleton for activities such as level walking, sit-to-stand transition or stairs climbing based on the literature. Key challenges and design to address them in a compact lower limb actuated exoskeleton with six DOFs are given. The mechanical design of AUTONOMYO has been presented and the device has been characterized without load at different gait cadences. This has clearly pointed out that AUTONOMYO fits high dynamics while providing important remaining assistance level, at least 50% to rise from a chair and about 75% to walk and climb stairs. Two points can still be improved however are the maximum velocity and peak torques at hip and knee flexion/extension.

ACKNOWLEDGMENT

Authors want to especially acknowledge Dohan Schlichtig, Marc Jeanneret, and Ortho Reha Wallner for their contribution to the mechanical design, their continuous support to the project and manufacturing of the parts of the exoskeleton.

REFERENCES

[1] D. Blackwell, J. Lucas, and T. Clarke, "Summary health statistics for U.S. adults: National Health Interview Survey, 2012," National Center for Health Statistics, 2014.

[2] Y. Zhang, A.-M. Chapman, M. Plested, D. Jackson, and F. Purroy, "The Incidence, Prevalence, and Mortality of Stroke in France, Germany, Italy, Spain, the UK, and the US: A Literature Review," *Stroke Research and Treatment*, vol. 2012, p. e436125, Mar. 2012.

[3] World Health Organization, *Neurological Disorders: Public Health Challenges*. WHO Press, 2006.

[4] G. Rosati, "The prevalence of multiple sclerosis in the world: an update," *Neurol Sci*, vol. 22, no. 2, pp. 117–139, Apr. 2001.

[5] A. Singh, L. Tetreault, S. Kalsi-Ryan, A. Nouri, and M. G. Fehlings, "Global prevalence and incidence of traumatic spinal cord injury," *Clin Epidemiol*, vol. 6, pp. 309–331, Sep. 2014.

[6] A. E. H. Emery, "Population frequencies of inherited neuromuscular diseases—A world survey," *Neuromuscular Disorders*, vol. 1, no. 1, pp. 19–29, 1991.

[7] M. Vukobratovic, D. Hristic, and Z. Stojiljkovic, "Development of active anthropomorphic exoskeletons," *Med. & biol. Engng.*, vol. 12, no. 1, pp. 66–80, Jan. 1974.

[8] R. Robotics, "German Social Court Ruling Deems ReWalk Exoskeleton Medically Necessary as Medical Aid for Recipient with Spinal Cord Injury." [Online]. Available: http://www.prnewswire.com/news-releases/german-social-court-ruling-deems-rewalk-exoskeleton-medically-necessary-as-medical-aid-for-recipient-with-spinal-cord-injury-300308952.html. [Accessed: 09-Jan-2017].

[9] "Ekso Bionics Bestowed With CE Mark," *Marketwire*. [Online]. Available: http://www.marketwired.com/press-release/ekso-bionics-bestowed-with-ce-mark-1658229.htm. [Accessed: 09-Jan-2017].

[10] "TÜV Rheinland Issues EC certificate for Cyberdyne's Medical Robot Suit HAL® | jp | TÜV Rheinland." [Online]. Available: http://www.tuv.com/jp/japan/about_us_jp/press_2/news_1/news_contentjp_en_168321.html. [Accessed: 09-Jan-2017].

[11] A. Ortlieb, M. Bouri, and H. Bleuler, "AUTONOMYO: Design Challenges of Lower Limb Assistive Device for Elderly People, Multiple Sclerosis and Neuromuscular Diseases," in *Wearable Robotics: Challenges and Trends*, Springer, Cham, 2017, pp. 439–443.

[12] M. Schenkman, R. A. Berger, P. O. Riley, R. W. Mann, and W. A. Hodge, "Whole-body movements during rising to standing from sitting," *Phys Ther*, vol. 70, no. 10, pp. 638-648; discussion 648-651, Oct. 1990.

[13] M. K. Y. Mak, O. Levin, J. Mizrahi, and C. W. Y. Hui-Chan, "Joint torques during sit-to-stand in healthy subjects and people with Parkinson's disease," *Clinical Biomechanics*, vol. 18, no. 3, pp. 197–206, Mar. 2003.

[14] D. A. Winter, "Kinematic and kinetic patterns in human gait: Variability and compensating effects," *Human Movement Science*, vol. 3, no. 1–2, pp. 51–76, Mar. 1984.

[15] D. C. Kerrigan, L. W. Lee, J. J. Collins, P. O. Riley, and L. A. Lipsitz, "Reduced hip extension during walking: Healthy elderly and fallers versus young adults," *Archives of Physical Medicine and Rehabilitation*, vol. 82, no. 1, pp. 26–30, Jan. 2001.

[16] A. Roaas and G. B. J. Andersson, "Normal Range of Motion of the Hip, Knee and Ankle Joints in Male Subjects, 30–40 Years of Age," *Acta Orthopaedica Scandinavica*, vol. 53, no. 2, pp. 205–208, Jan. 1982.

[17] S. Ounpuu, "The biomechanics of walking and running," *Clin Sports Med*, vol. 13, no. 4, pp. 843–863, Oct. 1994.

[18] A. Ortlieb, J. Olivier, M. Bouri, T. Kuntzer, and H. Bleuler, "From gait measurements to design of assistive orthoses for people with neuromuscular diseases," presented at the ICORR, Singapore, 2015.

[19] A. G. Schache and R. Baker, "On the expression of joint moments during gait," *Gait & Posture*, vol. 25, no. 3, pp. 440–452, Mar. 2007.

[20] J. Olivier, A. Ortlieb, M. Bouri, and H. Bleuler, "Mechanisms for actuated assistive hip orthoses," *Robotics and Autonomous Systems*, 2014.

[21] A. Protopapadaki, W. I. Drechsler, M. C. Cramp, F. J. Coutts, and O. M. Scott, "Hip, knee, ankle kinematics and kinetics during stair ascent and descent in healthy young individuals," *Clinical Biomechanics*, vol. 22, no. 2, pp. 203–210, Feb. 2007.

[22] M. R. Tucker *et al.*, "Control strategies for active lower extremity prosthetics and orthotics: a review," *Journal of NeuroEngineering and Rehabilitation*, vol. 12, p. 1, 2015.

[23] R. Baud, A. Ortlieb, J. Olivier, M. Bouri, and H. Bleuler, "HiBSO hip exoskeleton: Toward a wearable and autonomous design," presented at the MESROB, Graz, Austria, 2016.

[24] S. Rietdyk, A. E. Patla, D. A. Winter, M. G. Ishac, and C. E. Little, "Balance recovery from medio-lateral perturbations of the upper body during standing," *Journal of Biomechanics*, vol. 32, no. 11, pp. 1149–1158, Nov. 1999.

[25] R. W. Bohannon, "Comfortable and maximum walking speed of adults aged 20—79 years: reference values and determinants," *Age Ageing*, vol. 26, no. 1, pp. 15–19, Jan. 1997.

[26] G. Stoquart, C. Detrembleur, and T. Lejeune, "Effect of speed on kinematic, kinetic, electromyographic and energetic reference values during treadmill walking," *Neurophysiologie Clinique/Clinical Neurophysiology*, vol. 38, no. 2, pp. 105–116, avril 2008.