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TWIICE – A Lightweight Lower-limb Exoskeleton for Complete Paraplegics

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Abstract— **This paper introduces TWIICE, a lower-limb exoskeleton that enables people suffering from complete paraplegia to stand up and walk again. TWIICE provides complete mobilization of the lower-limbs, which is a first step toward enabling the user to regain independence in activities of the daily living. The tasks it can perform include level and inclined walking (up to 20° slope), stairs ascent and descent, sitting on a seat, and standing up. Participation in the world's first Cybathlon (Zurich, 2016) demonstrated good performance at these demanding tasks. In this paper, we describe the implementation details of the device and comment on preliminary results from a single user case study.**

I. INTRODUCTION

Each year, an estimated 250'000 to 500'000 people worldwide suffer a spinal cord injury, often leading to motor-complete paraplegia [1]. Loss of mobility and independence are part of the main consequences of paraplegia. Other health-related consequences include pressure sores, deficient bowel function and blood circulation, and loss of bone density [2]–[4].

Emerging from the field of robotics, powered exoskeletons have come as a solution to enable paraplegic people to walk again. Indeed, many projects are attempting to address the same problem. Among them, a few are already commercially available products, but most of them are still in the research phase. Each research team has adopted different strategies to overcome the major challenges residing in the design and implementation of a lower-limb exoskeleton and this has led to a wide diversity in the realizations. In particular, the following aspects are subject to important variations between the different developments: the number of actuated degrees of freedom and their actuation technology (type of motor and reducer); the materials used for the mechanical structure; the control strategies and how the user intention is detected. These choices have an impact on the overall device performance, which includes: ease of adoption by the user; physical burden exerted on the user; required level of external assistance before and during operation; total weight, bulkiness, operating time, and eventually cost of the device. A possible reason for the large diversity of exoskeleton designs is the difficulty to predict user and robot interactions. Both from a mechanical and behavioral standpoint, these interactions are critical for a good performance, but complex and highly subject-dependent. This challenge is thought to have led to the apparent "trial and error" strategy adopted by most of the research teams. It is yet another challenge to determine whether a given feature is beneficial for the use of exoskeletons. In general, a large scale clinical study is required before one can conclude on the influence of any design parameter on the final, application-related performance.

For an in-depth analysis of these features and how they vary between designs, the reader is referred to [5]. This review focuses only on devices that provide full mobilization of the lower limb. Other orthotic devices, such as partial orthoses, hybrid devices combining muscle functional electrical stimulation with actuators, and other anthropomorphic rehabilitation devices are not included in this study. For an overview of lower-limb assistive technologies, see [6]–[12]

Fig. 1: TWIICE, front and rear view. From top to bottom: electronics enclosure (1), back structure (2), hip actuation (3), thigh segment (4), knee actuation (5), shank segment (6) and foot plate (7). Cuffs with hook-and-loop straps (8) are used to create the mechanical interface with the user, in-between the legs.

Of note among exoskeletons for full mobilization are a few commercial products, already available for purchase. Those include Ekso GT (Ekso Bionics, Richmond, CA, USA) ReWalk (ReWalk Robotics, Yokneam, Israel) Rex (Rex Bionics, Auckland, New Zealand) and Indego [13]–[15] (Parker Hannifin Corporation, Macedonia, OH, USA). Because they are now commercial products, there is often limited information available about these devices. Ekso is designed for use in a clinical setting only. It comprises two

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active degrees of freedom per leg, at the hips and knees, in a longitudinal arrangement. Its ankle has a passive dorsiflexion joint, with a spring effect. The spring's equilibrium position can be adjusted. Hip abduction can be set to different fixed positions or set completely free. ReWalk was the first exoskeleton to be approved for personal use at home. It also has two active joints per leg, but no hip abduction. Its control is based on the user's posture, and so steps are triggered automatically. ReWalk is among the exoskeletons able to climb stairs. Indego is, to our knowledge, the lightest of all exoskeletons for complete paraplegia, but is not suited for stair ascent. Its lightweight fiber reinforced polymer structure, combined with a piecewise assembly, makes donning and doffing easy. Finally, Rex is remarkable in that it is self stabilizing without requiring crutches. This frees the user's hands, but comes at the cost of an extremely low walking speed and an increased overall weight.

On the academic side, some designs present some interesting features, which could have a significant impact on the devices' performance. Because of the pace at which the field of exoskeletons is evolving, some of the development mentioned here are not yet subject to publication. The Mindwalker, developed at the University of Twente, uses series elastic actuators (SEAs) as well an actively controlled hip ad/adduction, for weight shift and lateral foot placement [16]. Actuation of each joint is made through an outrunner BLDC motor and a ballscrew, connected to the distal segment via a spiral spring as series elastic element. Florida Institute for Human and Machine Cognition's exoskeleton for mobilization is called Mina [17], [18]. It provides actuation at both knees and hip, and has a fixed ankle joint with a compliant footplate. To date, no publication is available about Mina 2, the new version of the device, which comprises three degrees of freedom (DOFs) per leg instead of two. From the Walk Again project, the so-called Santos Dumont exoskeleton is remarkable in that it can be brain-controlled using EEG [19], [20]. It is also statically stable, so that the user does not need to ensure balance using crutches. Emerging from the Rehabilitation Engineering Laboratory at ETH Zurich, the VariLeg uses a variable impedance actuation at the knee joint. This enables adapting to environmental changes and uncertainty, but is also thought to better mimic the dynamic behavior of gait.

Finally, early-stage commercial products appear regularly, but very limited technical details can be found about them. Phoenix (US Bionics, Berkeley, CA, USA) has only one powered DOF per leg (the hip) while the knee uses a brake mechanism to take advantage of the passive dynamic properties of the knee joint. Its weight is close to that of the Indego (12.5 kg) but it does not enable to climb stairs either. From the company Bionik Laboratories (Toronto, Canada) Arke is designed for both rehabilitation setting and home use. Finally, Atalante by Wandercraft (Paris, France) is targeting self-stabilizing dynamic walking, but, to the best of our knowledge, this feature is still under test. With a total of 12 active joints, it provides power for almost all natural DOFs of the legs.

In this paper, we introduce TWIICE, a lower-limb powered orthosis. Among other features, TWIICE enables complete paraplegic users to get up from a chair, walk and climb stairs or slopes. With the help of forearm crutches, the user is in charge of his balance, turning, as well as weightshifting. He or she operates the device via buttons mounted on one of the crutches.

This paper makes a single user case report, six months after the first training session with a motor-complete paraplegic subject. We first detail the mechanical implementation of the device, including the design hypotheses, the actuation, the sensors and the electronics. Then the control architecture of the exoskeleton is presented. Finally, the device's performance is reported, as measured during preliminary assessment with one user. Among the presented results is the performance at the "Powered exoskeletons race" category of the Cybathlon 2016, to which TWIICE took part [21].

II. MECANICAL DESIGN

TWIICE's mechanical design was driven by one major rule: simplicity. The key idea behind it was to reduce the number and complexity of features, for the benefit of reliability and lightness. This strategy turned out to be a good one, as the device fulfilled expectations only 18 months after the beginning of the project. As a result of this robustness, the current prototype accumulated more than 44 hours of use with a paraplegic patient, and more than 10'000 footsteps without significant maintenance operation or reported mechanical failure.

TWIICE comprises four active DOFs in total. That is, only the hip and knee joints of each leg are actuated. All other DOFs are locked. The device mimics the kinematics of the human leg, having its hip and knee joints aligned with the user's joints. To compensate for the locked ankle joint, the foot sole is made with a special shape so that it can roll over the floor.

The leg's structure is made of carbon-fiber composite, significantly reducing the overall weight of the system. To date, two versions of the device's thigh and shank segments exist: an adjustable-length and fixed-length version. The former can be adjusted so as to fit several patients ranging from 135 cm to 155 cm in height, while the latter was tailored-made to best fit our Cybathlon pilot, 158 cm tall.

The back structure, linking the two legs and holding the user's torso, is adjustable in width and depth. It supports the control enclosure. Two belts are used to maintain the user's torso upright: a pelvic belt with three straps, and a thoracic belt. The last one is adjustable in height, and only required for higher injury levels.

The user's legs are kept aligned with the exoskeleton by the means of foam-covered semi-rigid cuffs (Fig. 1). They surround 60 % of the thigh and shank circumference and are rigidly attached to the exoskeleton's structure. Hook-and-loop fasteners maintain the user's legs firmly attached to the segments of the exoskeleton. Thanks to their flexibility, the cuffs can fit a wide range of morphologies, but they could be replaced easily if needed.

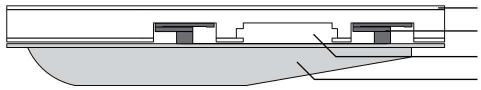


Fig. 2: Load cells assembly (sagittal cut view, only two of the four load cells are visible). The right side of the drawing corresponds to the front of the exoskeleton.

Characteristic	Value
Continuous torque:	34 N·m *
Peak torque:	57 N·m *
Weight:	1.4 kg
Back-driving torque:	12 N.m
Total reduction ratio:	140
Maximum tilting moment:	91 N.m

* Limited by the harmonic drive.

TABLE I: Characteristics of the actuation module

A. Actuation

The actuation module provides position control in revolution around the joint axis, as well as bearing of axial and radial loads and tilting moments. Torque is generated by a brushless DC motor. The current design features a modified version of the type SBZ-5612 with augmented torque constant and altered output shaft (Sonceboz SA, Switzerland). It is first amplified by a Gates timing belt stage with a 1:1.4 reduction ratio (Gates Corporation, USA). The belt then drives the input shaft of a Harmonic Drive (HD) CSD-20-2UH unit (Harmonic Drive AG, Germany). The HD unit has a reduction factor of 1:100 and a peak output torque of 57 N·m. It includes an output cross-roller bearing, supporting all loads and tilting moments applied to the joint. The general capabilities of the actuation module are summarized in table I.

B. Feet

Because of the fixed ankle joint, the foot sole must compensate in order to approximate a natural gait. The special rounded shape of the sole helps foreaft movements of the body as well as lateral weightshift during stance phases. The feet are fully rigid and do not bend when walking. The bottom face of the sole is covered by a rubber layer, in order to avoid slipping on smooth floors.

C. Electronics

The main parts of the control architecture are the on-board computer and the power electronics (fig. 3). They are located in a rigid, closed, and flat enclosure, in order to protect them.

The on-board computer is built around a BeagleBone Black (Texas Instruments, USA), a low-power single-board computer (1 GHz ARM processor, 512 MB RAM) running Linux. The program running on this board acquires the data from the sensors, manages the tasks' logic, and generates the joints trajectories. Data is continuously logged during operation, and stored on a micro-SD card. A custom daughterboard provides power supply, a 6-axes IMU, and a loudspeaker driver. A Wi-Fi dongle allows wireless communication with external devices, such as laptops, tablets or smartphones. The user input device is directly tethered to the on-board computer, in order to provide high reliability and minimal latency. The power electronics are custom PCBs designed to drive the brushless motors. They will herein be called ”motorboards”. They embed a STM32F7 microcontroller (STMicroelectronics, Switzerland), featuring a 216 MHz ARM processor with a floating-point computation unit. Each board controls two motors, so two boards are needed to drive the four motors of the exoskeleton. The communication with the on-board computer is made through a RS-422 differential serial bus.

An emergency stop button, located on the right side of the backpack (Fig. 1), can unpower all the motors, while keeping the sensing and the on-board computer functional. The angular position of each joint is sensed by the sine-cosine encoders on the motors. In addition, soft potentiometers (Spectra Symbol, USA) are installed on the joints, to initialize the incremental encoders, and provide redundant sensing for safety. In each foot sole, four flat load cells measure the vertical interaction forces of the foot with the floor (Fig. 2). Being the only mechanical link between the two rigid layers, they can measure accurately the weight on each foot. A small custom PCB amplifies the load cells signals, samples them, and sends the data to the on-board computer through a differential Serial Peripheral Interface (SPI) link. The embedded electronics have been designed to be modular, and can control other exoskeletons, such as the HiBSO hip orthosis described in [22]. The whole system is powered by a 48V, 220 W.h, 1.2kg lithium-polymer battery. Depending on the actions performed with the exoskeleton, it can last 2 to 3 hours.

III. CONTROL

A. User interface

After installation into the exoskeleton, the user can operate the device without the need of any external assistance. Remote control by a supervisor is also possible, which is useful during the first training sessions.

The user interface was designed to be as simple as possible, in order to minimize the learning period and decrease the mental effort. The user controls the exoskeleton with a dedicated input device fixed at the front end of the right crutch (fig. 4). This location allows him to access the three buttons with the thumb, and the trigger with the index finger, while still holding the crutch. A smartphone application displays information such as the current operation mode, or the remaining capacity of the battery. The user does not have to physically interact with it during normal operation, it only acts as a display. The loudspeaker located in the backpack provides acoustic feedback when interacting with the remote.

TWIICE is controlled with a system of action modes. Depending on the selected mode, pressing the trigger executes a different action. When no mode is selected, the exoskeleton stands still vertically. The action modes currently implemented are the following:

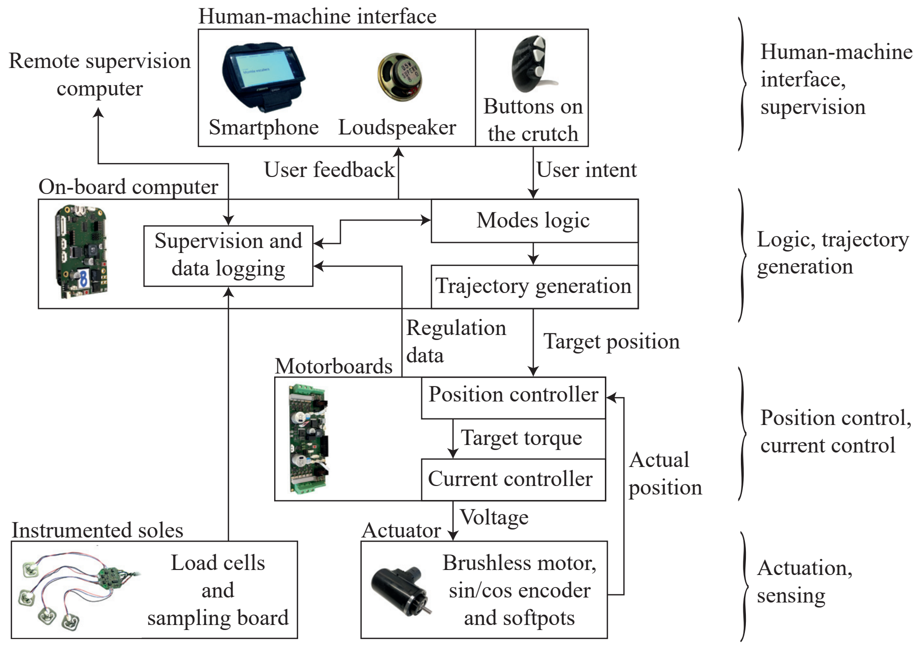


Fig. 3: Architecture diagram of the exoskeleton controller, including pictures of the most important electronic components. The blocks are laid out to emphasize the layered control structure. The arrows represent an abstracted information flow.

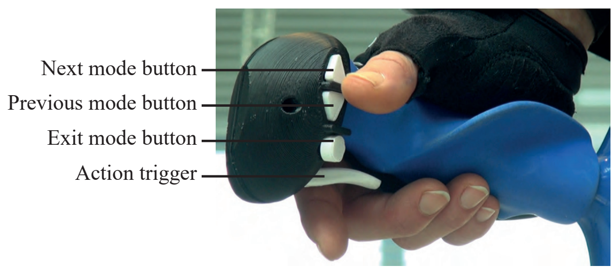


Fig. 4: Crutch remote. Thanks to push-buttons on the crutch handle, the user can trigger steps and change between operating modes.

1) Slow gait: each press of the trigger initiates a step. These steps are short and slow, which makes them suitable to approach an obstacle on the floor without kicking it.
2) Fast gait: each press of the trigger initiates a step. These steps are longer and faster, in order to travel at a higher speed.
3) Sitting down: when entering this mode, the exoskeleton gradually transition to the sitting position. Pressing the trigger will start standing up slowly. Meanwhile, the trigger can be pressed again to abort standing-up and sitting down again.
4) Stairs ascent: each press of the trigger initiates a step. Trajectories are designed to climb common stairs (17 to 19 cm step height).
5) Stairs descent: each press of the trigger initiates a step. Trajectories are designed to descend common stairs, backward.
6) Variable step length: each step is initiated by pressing the trigger, and the legs slowly keep spreading until the trigger is pressed a second time, or if the maximum step length has been reached. A third press on the trigger lifts the rear foot and brings the user back in the vertical position, so it can rotate on its front foot, and prepare for the next step. This mode was created specifically to overcome the "stones" obstacle of the Cybathlon [23].

To enter a mode, the user should first select it using the "next" and "previous" buttons, then activate it by pressing the trigger. To exit the mode, the "exit" button should be pressed. Each mode defines movements to transition smoothly in and out.

Before each step, the controller reads the feet load cells, to check that the body weight is supported by the stance leg. The step cannot be triggered until this condition is validated. The load cells and the IMU are not used yet to actively control the exoskeleton, but this will be the case in the future, when more sophisticated controllers will be implemented. Possible applications are adapting the steps to the terrain topology by sensing the foot contact, or triggering automatically the steps without the need to press the trigger.

B. Implementation

All the modes logic and the joints target trajectories are managed by the on-board computer. The target position samples are generated at 500 Hz. For all the modes, except the "variable step length", the joints trajectories are pre-defined off-line, and result in the same movement every time they are triggered. They are defined by piecewise cubic splines.

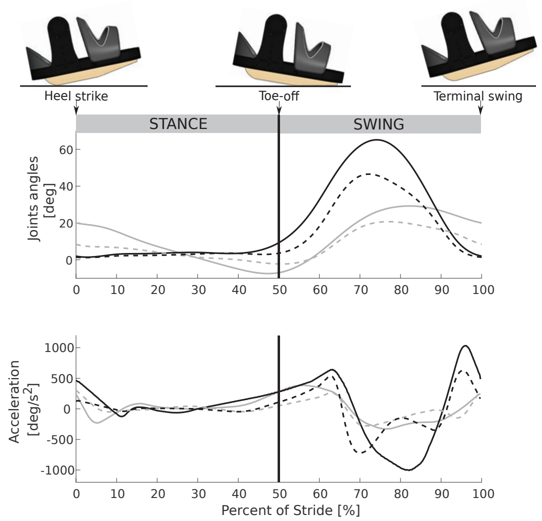
The control of the joints position is performed by the motorboards, with one proportional-derivative (PD) controller

per joint at a rate of 1 kHz. The computed motor torque is obtained by regulating the motor current with a proportional-integral (PI) controller at 10 kHz (fig. 3).
The board takes advantage of the redundant position sensors (incremental encoders and potentiometers, see section II-C) in order to detect a possible discrepancy between them, and trigger automatically the emergency stop. The emergency stop is also activated if the position regulation error is too large, or if the communication with the on-board computer is interrupted.

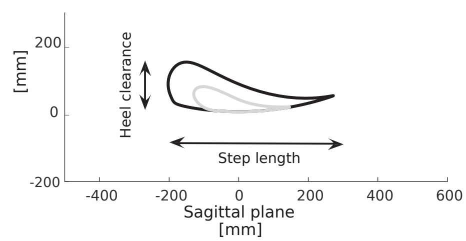
C. Trajectories

Although all the trajectories have been designed to match the human gait, some trade-offs had to be done due to the missing degrees of freedom at the hip and the ankle. For walking, the gait patterns (fig.5a) have less dynamical components than human walk on the ground at a medium cadence [24]. The damping performed by the knee after the heel strike has been removed in order to increase the stability of the pilot during the stance phase. The lack of mobility of the ankle during this phase is overcome by using the passive dynamics properties of a rocker-based inverted pendulum thanks to the curved foot soles. Moreover the legs are not moving during a short time between the stance and the swing phase; this break is needed by the user to adjust the position of the crutches, and ends when the trigger is pulled again to start the next step. A typical duration for this dead time is $0.7\,\mathrm{s}(\pm 0.1\,\mathrm{s})$ for a trained pilot. At the end of the stance phase, before the toe-off, the knee joint is bent slightly (9° for fast gait and 3.5° for slow gait) to mimic the push-off phase by being on the front part of the sole. During the swing phase, the knee flexion and the hip extension are higher than in the human gait to ensure a sufficient foot clearance due to the locked ankle.
The heel clearance is 142 mm and 72 mm for the fast gait and the slow gait, respectively (fig. 5b). The foot clearance of the fast gait is high enough to deal with uneven grounds. The fast gait has a step length of 450 mm, which is almost twice longer than the one of the slow gait (230 mm).

The stairs ascent is performed two at a time. Therefore, the stairs trajectories (fig.6) have been designed on the base of healthy individuals data [25]. As opposed to the human kinematics which keep the hip slightly flexed at the end of the stance phase, the hip extends until its zero position in order to keep the center of mass aligned with the foot in the sagittal plane. A trained user usually takes a break of approximately $1.5\,\mathrm{s}(\pm 0.3\,\mathrm{s})$ between the stance and the swing phase to move her crutch on the upper step and slide her hand along the handrail. During the swing phase, the knee joint stays flexed longer than in human kinematics, because a higher step clearance is needed due to locked ankle. At the end of the swing phase, just before contact, the foot is kept horizontal by setting the angles of the hip and the knee to the same value.
The stair descent is also done two at a time and uses the inverse trajectories of the stair ascent. Note that the user goes down the stairs backward due to the ankles limitations.



(a) Joints kinematics of the hip (grey) and the knee (black) and their accelerations for fast and slow gait, solid and dashed lines respectively, represented as a function of stride. The vertical line represents a pause in the gait cycle; corresponding to the time needed by the pilot to move the crutches forward.



(b) Foot locus of the heel for the two types of gait. The foot clearance increases as the steps length augments.

Fig. 5: Gait pattern in the sagittal plane.

The gait and the stairs kinematics were implemented by taking care of limiting the magnitude of the joints acceleration. Those can be felt by the subject and may cause loss of balance. Therefore, the magnitude of the acceleration has to remain small to avoid reactions that could disturb the balance and the comfort of the pilot. The accelerations for the gaits are smaller than $\pm 1000\,^{\circ}\,\mathrm{s}^{-2}$ (for a cycle period of 2s for fast gait and 2.5 for slow gait), and smaller than $\pm 600\,^{\circ}\,\mathrm{s}^{-2}$ for the stairs (cycle period of 3.57s for ascent and 6.6s for descent), see fig. 5 and fig. 6, respectively.

IV. ASSESSMENT METHODS AND RESULTS

Despite the increasing number of powered lower limb exoskeletons for complete paraplegic individuals, there is a lack of standardized quantitative metrics to assess the performance of those systems. Therefore, we chose to introduce the time needed to complete the obstacles of the Cybathlon 2016 as a set of standardized metrics to assess the efficiency of lower limb exoskeletons for daily living activities [23]. The main advantage of these obstacles is that their dimensions are well documented and thus are easily replicable. Our timing data come from the Cybathlon race, except for the

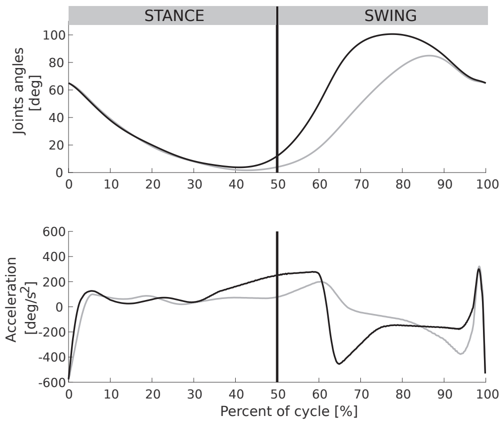


Fig. 6: Joints kinematics for stairs ascent.The joints angles of the hip (grey) and the knee (black), and their acceleration are presented as a function of cycle percentage. The vertical line represents a pause in the cycle allowing the pilot to adjust her upper body position

Obstacles	Time [sec]
Sofa	79
Slalom	33
Ramp and Door	132
Stones	NA
Tilted Path	NA
Stairs	115

TABLE II: Cybathlon's Results.

Tasks	Results
Walking speed	0.36 m/s
10-MWT	29 sec
Cadence fast gait	43 steps/min
Cadence ascent stairs	17 stairs/min
Cadence descent stairs	11 stairs/min

TABLE III: Gait performance

stairs. The tasks were timed line-to-line including therefore the modes transition and the positioning of the pilot with respect to obstacles (Table II). Since the stairs were not performed during the Cybathlon due to a technical issue, we used the recorded time during a training session prior to the competition. The obstacles "Stones" and "Tilted path" were not timed since the pilot was not able to complete them without assistance.

In addition to those metrics, we added the walking speed, 10-meter walk test (10-MWT), cadence of fast gait, and cadence of stairs ascent and descent.
The walking speed was calculated from the 10-MWT, since short-distance reflects lower extremity function [26]. For the 10-MWT, the user walked in a straight line in a hallway. The subject started walking several steps before the starting line that was known only by the recorder and walked through the second line. The subject was instructed to walk at a comfortable speed. The starting time and the final time were recorded when the subject's pelvic crossed the lines. The gait cadence was measured over 3 minutes of walking at a comfortable pace. The stairs cadence was measured over the mean of three ascents and descents of 22 steps each. These assessments were done on a single paraplegic subject with a T10 complete motor and sensory injury (ASIA A), 9 years post-injury. The subject is a 43-year-old woman, 158 cm (5.18 ft) tall and weights 45 kg (100 lbs). At the time of the Cybathlon, the subject had trained during 22 sessions of 2 hours, over 3 months. For the other metrics, the subjects had only 5 supplementary trainings of one hour. The subject had no prior experience of verticalization nor walking with Knee-Ankle-Foot-Orthosis (KAFOs). During testing, the subject was closely monitored by three spotters, although no assistance was provided.

The results of the Cybathlon's obstacles are shown in table II. In fast gait mode, the subject can currently reach a speed of 0.36 m/s (1.3 km/h) and has a cadence of 43 steps/min. The user climbs 17 stairs/min and descents 11 stairs/min (table III). Descending stairs takes more time than ascending them, since the movement is slower for safety reasons.

A. User experience

Each step is currently initiated manually using the trigger of the user input device. It may seem tedious, but actually the test pilot does not complain, because it makes the robot behavior very predictable.
The test pilot tends to excessively lean forward, and support a large part of their body weight with the crutches, because she is afraid of falling backward.

During the training sessions, we collected a feedback similar to [17] on the following points:

- A small step length makes weight shifting dramatically easier. Once the subject is trained, the step length can be increased gradually.
- It is difficult for the pilot to estimate the inclination of the whole body, on the sagittal plane. When leaning too far forward, this typically leads to hitting the floor with the tip of the foot, reducing the step size. Periodic oral hints from the assistance team, while walking, can mitigate this issue.
- Turning while walking is easily doable by the pilot when they are in single support phase. The fact that the surface in contact with the ground is small and hard helps reducing the friction. However, turning in place is performed by shifting repeatedly the weight on one leg then the other, while using the crutches to apply torque.

V. Discussion

The obstacles performance results have been measured only with a sportive test subject. The load applied to the crutches cannot be measured yet, so it may actually be too large, and not sustainable for other patients. A more comprehensive study should be performed later on. In particular, due to the fact that the ankle is locked, climbing stairs and a slope require significant contribution from the arms. In addition, the user currently cannot don or doff the exoskeleton alone.

Also, the pre-defined gait trajectories are fixed. In case of rough terrain, it is then difficult for the pilot to keep the lateral balance. Also, small terrain features can prevent the sole from rolling smoothly and destabilize the gait. Dynamic joints trajectories generation and a softer sole could solve the issue and will be tested in future test sessions.

VI. Future Work

In the next steps of the developments, TWIICE is expected to become more easily donned and doffed. It has so far been designed for participation to competition, which made the trade-off between performance and usability tend towards the former. But a better mechanical user-robot interface could enable the use of TWIICE independently. Among further improvements, a larger-scale version will be built, to fit users of up to 1m90 in height. Finally, the shape and the shock absorption of the curved sole will be improved to facilitate the weight transfer during walking and help the general balance.

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