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# Gait patterns generation based on basis functions interpolation for the TWIN lower-limb exoskeleton*

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*Abstract*— **Since the uprising of new biomedical orthotic devices, exoskeletons have been put in the spotlight for their possible use in rehabilitation. Even if these products might share some commonalities among them in terms of overall structure, degrees of freedom and possible actions, they quite often differ in their approach on how to generate a feasible, stable and comfortable gait trajectory pattern. This paper introduces three proposed trajectories that were generated by using a basis function interpolation method and by working closely with two major rehabilitation centers in Italy. The whole procedure has been focused on the concepts of a configurable walk for patients that suffer from spinal cord injuries. We tested the solutions on a group of healthy volunteers and on a spinal-cord injury patient with the use of the new TWIN exoskeleton developed at the Rehab Technologies Lab at the Italian Institute of Technology.**

## I. INTRODUCTION

A significant number of individuals in the world are affected by locomotion impairment due to traumatic events or medical conditions. The World Health Organisation statistics state that the total international incidence of spinal cord injury (SCI) is between 250 000 and 500 000 annual cases [1]. In most cases, repetitive and task-oriented movements of the impaired limbs can prevent complications such as muscle atrophy and osteoporosis [2]. For instance, numerous clinical studies support the effectiveness of an intense gait training, particularly in SCI and stroke patients [3]. However, the cost and the limited availability of assistance often prevent patients from achieving a satisfactory level of medical care in conventional therapy [4]. Training intensity may also be limited by fatigue of the therapist. Therefore, since the beginning of the $21^{st}$ century, researchers have been developing a vast range of robotic exoskeletons to actively move the users' impaired lower limbs through a series of motor activities, especially related to gait. Several experiments demonstrated that these devices have the potential to provide neurorehabilitation by moving the legs in a reproducible and cyclic way [5] [6]. Traditional designs use two degrees of freedom in each leg to obtain the flexion-extension of the knee and hip joints, which are generally actuated by electric motors. This is the case of the HAL [7], the Ekso [8] and the Indego [9] exoskeletons.

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Fig. 1: The TWIN exoskeleton.

The reference approach for controlling lower-limb exoskeletons in case of subjects almost or totally unable to voluntarily move their legs, like SCI patients, is the predefined gait trajectory method [10]. This control strategy consists in imposing a predefined gait trajectory, *i.e.* a sequence of desired joint angles in time, which is generally inspired from a healthy person and replicated by the exoskeleton. This control type is used, for instance, by the REWALK [11] and ATLAS [12] exoskeletons.

A common method to generate a gait pattern is to determine the joints positions with the help of predefined via-points and fit a mathematical curve to them [13]. Nevertheless, a well-known drawback of working in joint space is the difficulty in regulating the cartesian parameters of gait, such as step length and height [14].

In this work we present our novel approach of generating gait trajectories for SCI patients. This is the result of a tight cooperation with clinicians from two major rehabilitation clinics in Italy and it was validated by direct experimentation on a SCI patient. In Section II, an overall structural description of the lower-limb TWIN exoskeleton (Fig. 1) is presented. In Section III, our generation method is explained. Results of the experiments where we tested all our trajectories are presented in Section IV and discussed in Section V. Section VI shows our conclusions and future works.

## II. TWIN LOWER LIMB EXOSKELETON

TWIN is a powered lower limb exoskeleton developed at the IIT-INAIL Rehab Technologies Lab. Its overall structure consists of four actuated joints (at the hips and at the knees) connecting five rigid sections: one pelvis component and two separate, right and left, segments for thighs and shanks.

The actuation units employed to power all four joints are based on EC90 flat Maxon motors coupled with 100:1 gearboxes. The device requires the use of forearm crutches to maintain balance during standing and ambulation.

Fig. 2 shows the system architecture of the TWIN exoskeleton: a six point star client/server structure with 4 motors, 1 Inertial Measurement Unit (IMU) and 1 battery interface. The exoskeleton is operable via an Android device that employs a custom-made app that communicates via Bluetooth and has been developed following the IEC 62304 regulation for class A medical devices for safety reasons. The central control unit (CCU), located in the backpack of the device, is a custom board based on an ARM Cortex A9 processor which is in charge of monitoring the battery pack and the 6-axes IMU sensor that are connected to it via its I2C interfaces. Each motor is locally controlled and actuated in position via a proportional-integrative (PI) controller, operating at a frequency of 20 kHz, thanks to a logic board based on an ARM Cortex-M4 microcontroller unit. The communication between the CCU and the motor controllers is performed via CAN bus at a frequency of 500Hz.

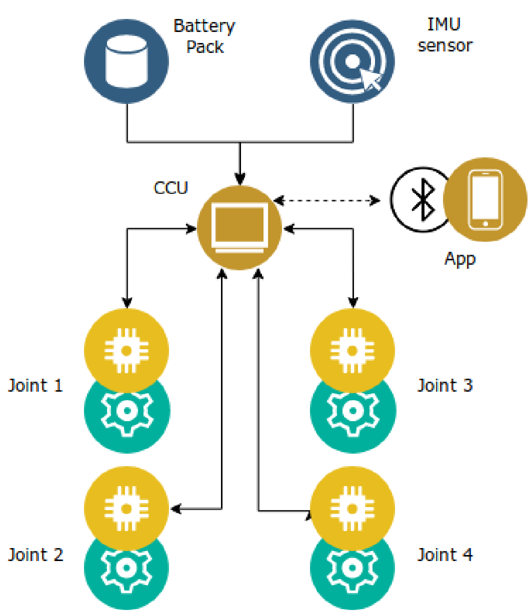


Fig. 2: Logical structure of the TWIN exoskeleton and main elements of the control software (in yellow)

## III. TRAJECTORIES COMPUTATION

The capability to generate acceptable, stable and effective gait patterns is one of the critical points in development and clinical use of lower-limb exoskeletons today. Among other requirements, these devices have to provide a stable walk to prevent any discomfort or risk of fall. Given the system described in Section II, the walking problem translates into the generation of sequences of setpoints position that, at each iteration ($f = 500Hz$), the CCU sends to the motor controllers.

### A. Computing the reference trajectory

In this paper we present three different walking styles based on three different trajectories: T1, T2 and T3. In order to generate them, we exploited an interpolation approach which consists in multiplying the length $L$ and the height $H$ of the desired step by some basis functions that are normalized both in amplitude and over time. Indeed, it is possible to obtain the reference trajectories $(x_F, z_F)$ and $(x_T, z_T)$ in Cartesian space as follows:

$$x_F(t) = x_F^0 + L * f_x(t) \tag{1}$$
$$z_F(t) = z_F^0 + H * f_z(t) \tag{2}$$
$$x_T(t) = x_T^0 + L * g_x(t) \tag{3}$$

where $x_F^0$, $z_F^0$, $x_T^0$ are the configuration of the swing foot and the torso at the beginning of the step, while $f_x$, $f_z$ and $g_x$ are the normalized basis functions. For each trajectory style T1, T2 and T3, they are shown in Fig. 4 and Fig. 5.

We decided to neglect the torso trajectory $z_T$, because it is not needed given the following assumption on the ankle passive joint: the flexion angle $\beta$ (shown in Fig. 3) varies depending on the step length *i.e.* the longer the step the greater the flexion. Assuming that: $\{\beta \in \mathbb{R}, \beta_{min} \leq \beta \leq \beta_{max}\}$ and $\{L \in \mathbb{R}, L_{min} \leq L \leq L_{flex}\}$, where $\beta_{min}$, $\beta_{max}$, $L_{min}$, $L_{max}$ are respectively the minimum and the maximum values of the ankle joint angle and the step length, we define $\beta_{flex}$ as the maximum flexion of the ankle during a step. In our trajectory design, the relation between $\beta_{flex}$ and $L$ is considered to be linear and it can be described as follows:

$$\beta_{flex} = \beta_{min} + (\beta_{max} - \beta_{min}) * \frac{L - L_{min}}{L_{max} - L_{min}} \tag{4}$$

$$\beta(t) = \beta_{min} + (\beta_{flex} - \beta_{min}) * g_x(t) \tag{5}$$

Considering that the movement of the leg in support has to be minimized, the ankle $\beta$ and the torso reference $x_T$ are multiplied by the same basis function $g_x$. Following this assumption, the resulting trajectory of the knee joint is around 0°, while the hip joint performs little movements. Note that $L_{min}$, $L_{max}$ are defined values, that depend on the structure of the exoskeleton: in our case we considered $L_{min} = 0.05m$ and $L_{max} = 70\%$ of the leg length.
Likewise, $\beta_{min}$ and $\beta_{max}$ are structural dependent values, their tuning is described in the following section.

### B. Geometric model

The mathematical model of the TWIN exoskeleton is based on the Denavit-Hartenberg method [15]. For each joint, the rotation axis is oriented such that the angle is considered negative when the leg is moving backwards. An illustration of the robot parameters is shown in Fig. 3: $q_1$ and $q_2$ are respectively the knee and hip angles of the leg in support, while $q_3$ and $q_4$ are the hip and the knee angles of the leg in swing. The tibia, femur, and hip dimensions are defined as $l_T, l_F$ and $l_H$ respectively. The angle $\theta_T$ is the angle of the torso.

The ankle joints are passive and consist of two springs that can be tuned to modify $\beta_{min}$ and $\beta_{max}$. The tuning takes in consideration the patient characteristics: his/her ankle flexibility, comfort during movements, and the ability to walk in a relaxed way. These are some of the key points that physiotherapists have to analyze in the training process, in order to correctly tune the stiffness of the ankle joints. The lower the stiffness, the higher the mobility: an inexpert patient might feel unstable if the ankles are too flexible, while skilled ones might feel limited and tied if they are too stiff.

The angles $\theta_T$ and $\theta_F$ are respectively the angles of the torso and the ankle of the swing leg w.r.t. the z-axis. We constrain the latter equal to zero, at each instant of time, in order to compute a trajectory such that the support leg is referenced againts a vertical torso. This approach aims to mitigate the SCI patient propensity to lean forward while walking. In our geometric model, the $\theta_F$ angle is negligible since in our approach we do not control the orientation of the swing foot.

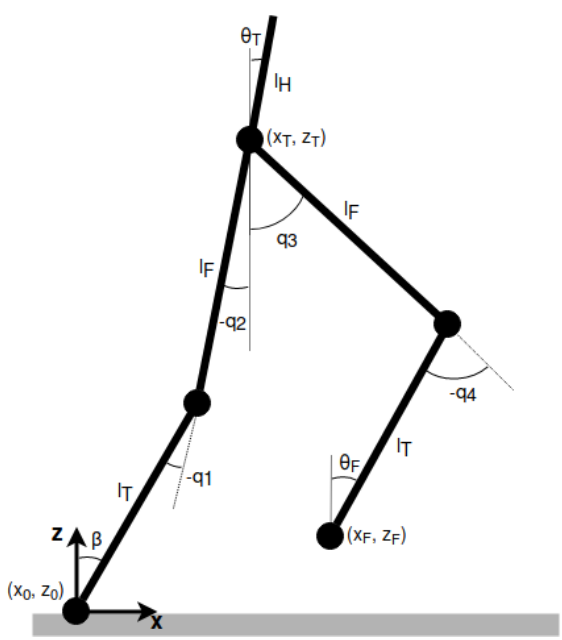


Fig. 3: The TWIN parameters description

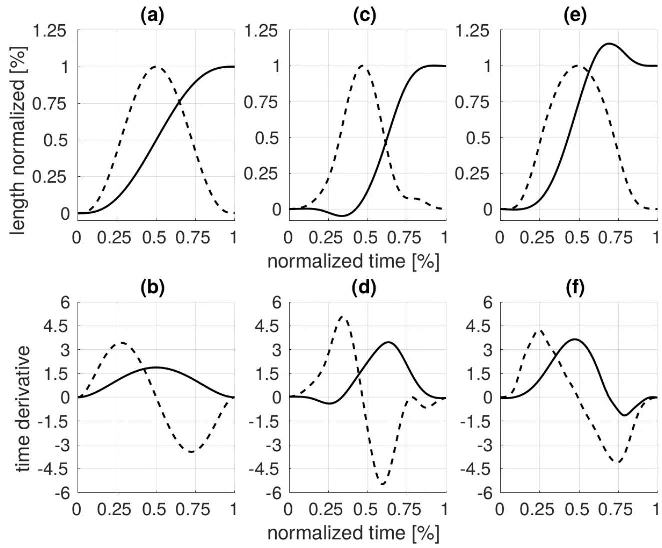


Fig. 4: The basis functions of trajectory T1, T2 and T3 (respectively (a), (c) and (e)), and their time derivatives (respectively (b), (d) and (f)). The basis functions $f_x$ are represented by continuous lines, while $f_z$ are represented by dotted ones.

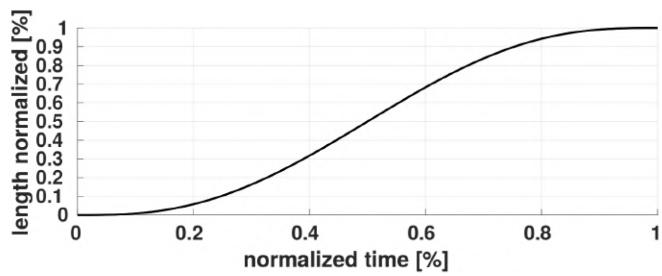


Fig. 5: The basis function $g_x$, polynomial of sixth-order, exploited for the generation of the reference trajectory $x_T$.

### C. Inverse Kinematics

In order to obtain the joint trajectories, we divided the inverse kinematics problem in two parts: the swing leg follows a reference trajectory $(x_F, z_F)$ while the support one obeys to the torso reference trajectory $(x_T, \theta_T)$. Given the following assumptions: (i) $\theta_T = \beta + q_1 + q_2$ with $\{\beta \in \mathbb{R}, \beta_{min} \le \beta \le \beta_{flex}\}$, (ii) $\theta_T(t) = 0, \forall t \in [0, t_s]$ where $t_s$ is the step duration, the kinematics equations are:

$$q_1 = sin^{-1}\left(\frac{x_T - (l_T \cdot sin(\beta) + l_H \cdot sin(\theta_T))}{l_F}\right) - \beta \quad (6)$$

$$q_2 = \theta_T - q_1 - \beta \quad (7)$$

$$q_3 = cos^{-1}\left(\frac{a_x^2 + a_z^2 - l_F^2 - l_T^2}{2 \cdot l_F \cdot l_T}\right) \quad (8)$$

$$q_4 = tan^{-1}\left(\frac{z_F - z_T}{x_F - x_T}\right) - tan^{-1}\left(\frac{l_T \cdot sin(q_3)}{l_F + l_T \cdot cos(q_3)}\right) \quad (9)$$

### D. Design of the basis functions

We worked closely with physiatrists and physical therapists, during the entire design process, to sequentially develop three different trajectories. At the beginning, a first trajectory - defined as T1 - was generated starting from the idea that a gait pattern should be first and foremost safe in order to minimize the risk of stumble or fall. A second trajectory T2 was successively designed adopting a curve profile that is similar to those exployed in currently available lower limb exoskeletons. Finally, the third trajectory T3 was created on the basis of the results obtained with T2 and some further adaptation in order to improve its safety by taking into account past studies on humanoid robots. Our employed approach also consisted in quick iterative cycles where we collected feedback from users and clinicians at every step to increasingly improve each trajectory and generate the next one. All the basis functions were optimized in terms of acceleration minimization, *i.e.* minimum jerk planning [16] [17], in order to obtain smooth joint trajectories.

The three obtained trajectories are described more in details in the following:

*1) Trajectory T1:* it has been designed keeping focus on the simplicity of movement. The step is executed in two distinct phases: the rising and the landing of the foot both last exactly half of the total swing time. Therefore the peak height is reached at $t_H = 0.5 * t_S$. In addition to this, the shape of the step has been designed to be symmetrical: $z_F(t_H - \epsilon) = z_F(t_H + \epsilon)\ \forall \epsilon \in [0, t_H]$. The resulting basis functions (Fig. 4 (a)) $f_x$ and $f_z$ as polynomials of $6^{th}$ and $7^{th}$ order respectively. A geometric representation of the trajectory T1 is illustrated in Fig. 6(b). The trajectory T1 was the first one to be implemented in the TWIN exoskeleton. Its design allowed us to compare a "simple" walking style w.r.t. more sophisticated ones.

*2) Trajectory T2:* the basis functions shown $f_x$ and $f_z$ in Fig. 4(c) have been designed in order to induce the foot to move backwards during the toe-off phase (*i.e.* when the foot leaves the ground, heel first, toe last), to reach the maximum height during the first half of the step and then to perform a plane-like landing. In recent years, the lower-limb exoskeletons TWIICE [18], Lokomat® [19] and C-ALEX [20] have used similar walking trajectories for rehabilitation therapies and paraplegic assisted walking. A geometric representation of the trajectory T2, implemented in our device, is shown in Fig. 6(d).

*3) Trajectory T3:* starting from T2, we designed a third trajectory shape in order to accentuate the heel-strike phase, *i.e.* when the foot touches the ground heel first. The importance of the heel-strike to toe-off movement has been widely studied in literature: it is believed that the arches, functioning as springs, are compressed in the early stance to store mechanical energy and then to convert it into propulsive power during toe-off. This anatomically passive mechanism reduces metabolic energy during locomotion and helps with the balance therefore improving its safety [21] [22]. Even in the humanoid robotics field, the importance of heel-strike to toe-off phase has been examined: the rolling movement heel-toe can increase the stride length compare to flat walking, allowing faster movements, reduced loading of the leg joints and improved stability [23] [24]. Considering the uncontrollable ankle joint, and the use of crutches to preserve balance, we designed two basis functions that emphasize the heel-strike movement. A representation of these functions is shown in Fig. 4(e), while the geometric foot clearance of T3 is shown in Fig. 6(f).

## IV. Assessment methods

### A. Participants

The assessment of the aforementioned gait patterns has been performed as a two-stage process: in the first phase a group of ten healthy volunteers participated in the experiment. They had no neurological or muscular pathology that would affect their locomotion. The focus of this stage was to acquire user's feedback throughout the 36 months period of the TWIN development phase; in this way the subjects could test, in time, all the trajectories T1, T2 and T3 while gaining expertise and a better walking capability. The second stage of the experimental assessment involved one 31-year-old SCI patient with a D5 lesion and significant previous experience in the usage of lower-limb exoskeletons. The focus of this stage was the evaluation of the feasibility of the proposed solutions with an impaired user.

All the participants gave written and informed consent before their inclusion in the study. The experiments respect the standards of the Declaration of Helsinki (rev. 2013), with formal approval of the ethics evaluation committee Comitato Etico Interaziendale Bologna-Imola of the Pharmaceutical Department U.O.C. Farmacia Ospedale Maggiore, Bologna, Italy (Protocol number: CP-POR1-01 ver.01).

### B. Protocol

The experiments took place in empty areas at least 15x6 meters wide under the supervision of qualified operators to ensure at all times the safety of the participants. All of them were instructed on how to use and behave inside the exoskeleton. Preliminary walking activities were also conducted to further acquaint the participants with this kind of device. We defined a starting point P1 and a final point P2 located ten meters away. The participants were asked to walk straight at their preferred speed from P1 to P2 more than once in each testing session (Fig. 7). Speed was selected online by using the Android application while the participant was standing still, not during walking, for safety reasons.

### C. User experience

The assessment phase provided qualitative feedback for the development of the three trajectories. With regard to trajectory T1, all the participants felt it uncomfortable due to the unnatural flexion of the knee (Fig. 7 (a)). On the other hand, assessing T2, the problem of stumbling on the ground with the toe was commonly faced (Fig. 7 (b)). We associate this issue to the lack of degrees of freedom in the

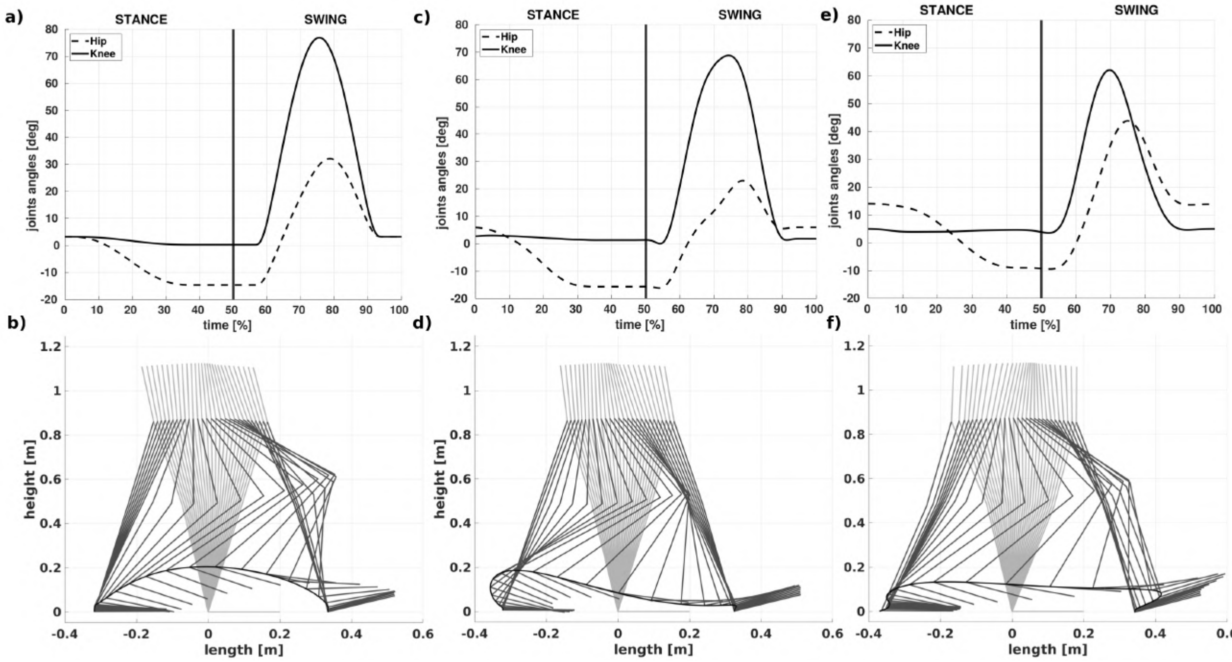


Fig. 6: The reference joint trajectories, in position control, during a stride are shown on the first row, while the geometric representation of the swing foot during a step is shown on the second row. In all the cases, the step is ∼0.40m long with a peak height of ∼0.20m. (a) and (b) show trajectory T1, (c) and (d) show trajectory T2, (e) and (f) show trajectory T3.

ankle. In order to avoid the problem, the participants had to tilt backwards the torso making the walk uncomfortable and unstable. Lastly, T3 was considered unanimously the smoothest trajectory without presenting any problem related to ground collision (Fig. 7 (c)).

## V. Discussion

The basis function interpolation provides an easy-to-use and reliable tool to generate customized gait patterns for a lower-limb exoskeleton. Given the mathematical model of the robotic device and the geometric properties of the desired gait, the trajectory components on the sagittal plane can be analyzed in order to identify the basis functions able to reproduce the desired patterns with minimum jerk planning. In this section the outcomes of the experimental process are presented, highlighting the positive and negative aspects encountered for each gait style T1, T2 and T3.

One representation of the trajectory T1 in the sagittal plane is shown in Fig. 6 (b). The resulting evolution of the knee and the hip joint angles during stance and swing phases is shown in Fig. 6 (a): it is possible to observe that, during the swing phase, the knee joint performs a significant movement while the hip joint plays a secondary role. Indeed, the former reaches a maximum angle of $\sim 75°$ while the latter has a peak of $\sim 30°$. Their movements are synchronized and they reach the maximum excursion almost at the same time. This wide range of motion explains why all the participants felt uncomfortable as the excessive knee flexion, combined with a much lower hip flexion, created a sense of instability to the patient as it did not exploit the singularity which could be offered by his skeletal system if fully extended. We conclude that T1 results to be functional, without stumble problems, but unnatural and uncomfortable.

The geometric representation of T2 is shown in Fig. 6 (d), while the resulting knee and hip joint trajectories are depicted in Fig. 6 (c). It results that the former reaches a maximum knee flexion angle of $\sim 70°$ ($\sim 5°$ less than T1) while the latter has a peak of $\sim 22°$ during the swing phase. Their maximum excursions are almost synchronized around the 70-80% of the stride. The gait style T2 allows to perform walking steps with a much smaller excursion variation of the joints angles w.r.t. T1. However, although this trajectory minimizes the patient's effort and it is much more human-like than T1, the participants in some cases faced stumbling problems. An accurate tuning process of walking parameters was required in order to minimize the risk of collisions with the ground.

The trajectory T3 has been developed in order to obtain a pronounced movement of heel-strike that implicates a propulsive power during toe-off phase. Moreover, we wanted to identify basis functions able to generate a gait cycle which is comfortable and stable for healthy and SCI subjects, without requiring an accurate tuning of walking parameters. The resulting trajectory is able to minimize the joint angles variation and avoid the stumbling problem. We identified the basis functions $f_x$ and $f_z$, shown in Fig. 4 (e-f), such that the toe-off phase is defined by the foot that moves up and forward at the same time, then goes parallel to the ground up to $110\%$ of the step length, and finally ends landing back

(a)

(b)

(c)

Fig. 7: Snapshots of the experiments (a) SCI patient testing the T1 trajectory at the Centro Protesi INAIL premises. (b) Healthy subject stumbling on the ground while testing the T2 trajectory at the Rehab Technologies Lab premises. (c) Healthy subject testing the T3 trajectory at the Rehab Technologies Lab premises.

at 100%. Thanks to this behavior, the heel-strike phase is well pronounced, providing momentum to the user for the next step. Fig. 6 (e-f) shows the joints trajectories and the shape of the trajectory T3. We can observe that the excursion between hip and knee joint angles is considerably reduced w.r.t. the trajectories T1 and T2: the former has a peak of $\sim 60°$ and the latter of $\sim 42°$. The contribution of the hip joint is higher in T3 than in previous trajectories, reducing the knee joint angle variation. Such synergy leads to have a more comfortable and smooth and safe gait pattern w.r.t. T1 and T2 and therefore represents the trajectory that satisfies the set requirements the best.

The gait styles T1, T2 and T3 have been also tested with the SCI patient which has been able to walk successfully with the proposed trajectories. The outcomes of this activity substantially confirmed the feedback obtained in the development phase.

## VI. Conclusions and future work

In this paper we presented three different gait patterns: T1, T2 and T3. In order to generate them, we exploited an interpolation approach multiplying normalized basis functions by the geometric properties of the desired gaits. The design of the proposed gait patterns was done thanks to the cooperation of field experts and validated in a clinical environment on a SCI patient. The main conclusions that can be draft from this work are that an effective gait pattern should be able to *(i)* minimize the joint angles variation during walking to maximize the comfort of the user and to *(ii)* accentuate the heel-strike phase thus avoiding the stumbling problem and increasing the stability of the system. Therefore, we conclude that the trajectory T3 yields so far the best results in terms of effectiveness and acceptability and it could be a reference paradigm for future studies on lower limb exoskeletons. In the next step of the TWIN project, we plan to fully evaluate the acceptability of the proposed gait patterns on a statistically significant number of impaired patients.


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