On the Potential Field-based Control of the MIT-Skywalker

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***Abstract*— Walking impairments are a common sequela of neurological injury, severely affecting the quality of life of both adults and children. Gait therapy is the traditional approach to ameliorate the problem by re-training the nervous system and there have been some attempts to mechanize such approach. We have recently presented the MIT-Skywalker; a novel device to deliver gait therapy, which, in contrast to previous approaches, takes advantage of the concept of passive walkers and the natural dynamics of the lower extremity in order to deliver more "ecological" therapy. In this paper we present a control scheme for the MIT-Skywalker, which is based on an artificial potential field applied at the foot workspace. It is used to improve sensory feedback to the patient, as well as to increase to normal the range of motion of the paretic leg. Simulation results prove the efficiency of the proposed controller.**

I. Introduction

Every 40 seconds, someone in the United States has a stroke [1]. For every 1,000 children born in the US, 2.8 youngsters have cerebral palsy [2]. The impact of these and other neurological pathologies on walking is significant and locomotor capacity is a critical factor in determining an individual's degree of disability. Physical therapy is the standard of care to educate the individual on how to ameliorate or regain walking abilities. We introduced a paradigm shift in 1989 when we started the development of the MIT-Manus [3]. The goal was to provide robotic tools to facilitate and increase the productivity of clinicians while optimizing the potential for patients to recover.

Yet, to employ mechanical devices to deliver therapy is not a new idea with the most common mechanical device used in gait therapy being the treadmill. Treadmill training offers task-oriented repetitive movements that can improve muscular strength, aerobic capacity, and movement coordination [4]. Body weight support treadmill training (BWSTT) has been shown to improve gait and lower-limb motor function in patients with locomotor disorders [5], [6]. Weighted-supported treadmill training for hemiparetic patients has been shown to improve balance, lower-limb motor recovery, walking speed, endurance, and other important gait characteristics, such as symmetry and stride length [6]; presently there is a large NIH sponsored randomized clinical trial (RCT) study (www.leaps.usc.edu; Principal Investigator: Pamela Duncan).

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However, BWSTT requires a therapist to monitor and manipulate the pelvis in addition to one or two therapists needed to propel the leg(s) forward. Robotic devices were built in an attempt to automate the therapy process further. While several robotic devices already exist (e.g. to manipulate the ankle: MIT's Anklebot [7], to provide smart body weight support: KineAssist [8], Zero-G, to manipulate the pelvis: UC Irvine's Pam and Pogo [9], MIT-Pelvis robot, to manipulate the foot: Gait Trainer I, Haptic Walker, G.E.O, to manipulate the knee and hip: Lokomat, Lopes, Motorika/Healthsouth Autoambulator), presently only two devices have been used extensively with more than 20 patients with published clinical outcomes, namely the Gait Trainer I and the Lokomat. Gait Trainer I is an end-effector based robot with quick set-up time, incorporating both an adjustable Body Weight Support (BWS) and sliding foot plates that are secured to the patient's feet [10]. While it minimizes the number of therapists to only one needed to manipulate the knee, the planar sliding motion reproduces the kinematic but it does not reproduce heel-strike. The Gait Trainer I was tested in a large multi-site RCT, DEGAS study, with positive results. The Lokomat system is an exoskeletal device. It includes a treadmill, an active BWS (newest generation), and a robotic orthosis with four degrees of freedom, actuating left and right knee and hip joints [11]. This device attempts to replicate the kinematics of an unimpaired subject, but it does not incorporate any means to promote weight shifting from one leg to the other and also forces the ankle to be always in a dorsiflexed position. Although there were some positive pilot results using the Lokomat [12], more recent studies found that Lokomat training had no advantage compared to conventional therapy [13], [14]. Because of the much higher intensity of training in the Lokomat group, we interpreted this result as an indication that the Lokomat kinematic experience might not be affording the proper neurological stimulus.

The introduction of robotic devices in the therapy allows for the application of a wide range of rehabilitation programs and protocols, which should not be limited to the imposition of rigid kinematics. There have been some attempts in the past to create a performance-based scenario for the robotized therapy of the upper limb, using an impedance controller for making the therapy interactive [15]; a predefined desired trajectory for the hand was presented to the patient, while the robot applied corrective forces to the patient's hand, when the latter was either deviating from the desired trajectory or delaying to initiate movement. On the lower limb, Hocoma and the ETH Zurich initiated pilot testing on an improved version of their software that affords a more interactive experience [16], using a similarly defined impedance controller. However, focusing on the lower limb, following only a desired trajectory using robot's aid has been proved unsuccessful towards motor recovery [13]. Certain neurophysiology studies have shown that human locomotion is a dynamic process that is greatly affected by a variety of afferent systems and sensory information generated by the interaction of the foot with the environment [17], [18]. Therefore, a controller that would also provide the appropriate feedback to the patient leg could improve therapy's efficiency.

We have recently completed the alpha-prototype of the MIT-Skywalker. This novel rehabilitation robot is unique and distinct from other existing rehabilitation robotic devices for gait. It delivers safe and efficacious gait therapy inspired by the concept of passive walkers [19]. The MIT-Skywalker creates the required ground clearance for swing while exploiting gravity to assist during leg propulsion. Preliminary tests with a mannequin and unimpaired subjects, demonstrated its ability to allow gait therapy without restricting the movement to a rigid kinematic profile, providing ecological heel-strike and hip-extension and maximizing patient participation during therapy.

In this paper, we present a control scheme for the MIT-Skywalker that a) provides the necessary neural feedback to the patient by simulating "ecological" contact events (e.g. heel-strike), and b) helps towards the increase of the paretic's leg range of motion by adding energy to the system-leg when needed. The controller is realized through the introduction of an artificial potential field inside the foot workspace, which is applied through the unique non-interfering actuation mechanism of the MIT-Skywalker. Simulation results show the efficacy of the proposed method.

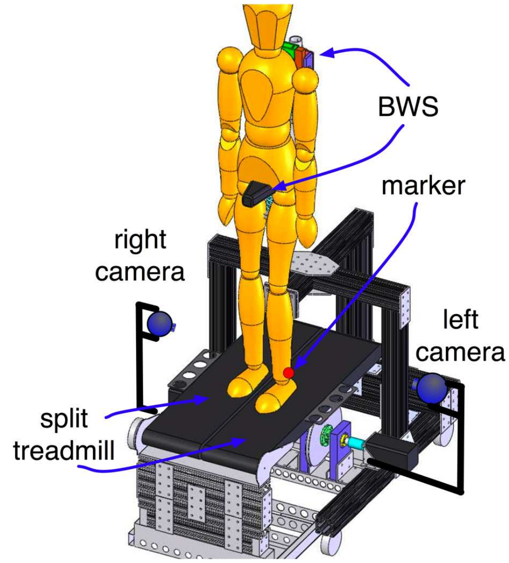

Fig. 1. The MIT-Skywalker platform. The Body Weight Support (BWS) is consisted of a saddle-like seat and a trunk support to stabilize the patient's torso. Two cameras and markers are used for foot position feedback.

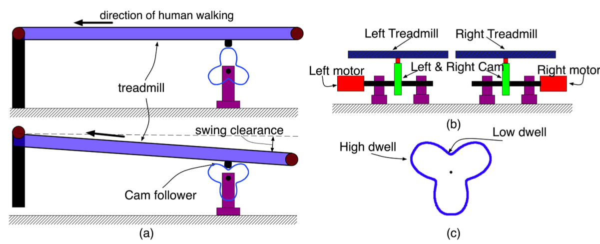

Fig. 2. (a) Side view of the cam-based vertical actuation system of the treadmill. A $60°$ rotation of the cam lowers the one end of the treadmill to provide the necessary swing clearance. (b) Front view of the motor-cam systems for each of the two belts of the split treadmill. (c) The cam profile. The high-low dwell sequence was replicated three times for allowing the use of the cam at future applications with continuous rotation of the cam.

II. Description and Control of the MIT-Skywalker

A. The MIT-Skywalker platform

In conventional gait physiotherapy, the therapist pushes or slides the patient's swing leg forward, either on the ground or on a treadmill. In clinic-tested robot-assisted gait therapy, the leg is propelled by either the robot orthosis acting on the patient's leg (Lokomat), or foot plates attached on the patient's foot (in Gait Trainer I). Instead of lifting the patient's leg manually or mechanically, we achieve forward propulsion in the MIT-Skywalker by lowering the walking surface. This provides both swing clearance and takes advantage of dynamics and gravity to propel the leg forward, while allowing proper neural inputs for hip extension and ecological heel strike.

Our alpha-prototype was presented in [20] and it is shown in Fig. 1. It consists of a split treadmill, which stays horizontal during the stance phase of each leg, and may be lowered to provide swing clearance for the impaired leg. The tracks of the split treadmill are lowered through a cam system, shown in Fig. 2a, 2b. Therefore the rotation of the cam controls the vertical position of each of the right and left treadmills. The rotation angle of the cam is directly controlled through a servo-controlled brushless motor. The cam high and low dwell were appropriately designed to allow for the necessary swing clearance, while the cam profile was designed using a continuous smooth function to avoid high acceleration, i.e. forces at the cam surface (Fig. 2c). Finally, the cam high-to-low and low-to-high profiles were replicated three times, in order to accommodate future versions of the treadmill actuating system using constant speed cam rotation.

The alpha-prototype also includes a body-weight support system, since many patients are not able to support their weight on the impaired leg(s) or they may need assistance maintaining balance. This system provides enough support to unload up to 100% of the patient's weight and keep the patient safe from falls, yet not interfere with the required ranges of leg motion. While kinematically-based devices employ overhead full-body harnesses, we designed a system to afford fast do-on and -off. It consists of a simple chest harness providing stabilization for the upper body and a saddle/bicycle-like seat for body-weight support [21]. The seat is mounted on a system of 4 springs of different stiffness values selected to position the pressure points on the subject buttock and 2 rotation joints that allow both vertical motion (max. displ. 1 in) and roll and pitch rotation (max. rot. $5°$).

The control of the cam system is key in providing the required swing clearance for the patient's leg and ecological hip-extension and heel-strike. Although we can probe the state of the device, we required feedback of the patient's leg in order to control the treadmill speed and cam system. The feedback system used is equipped with two cameras placed on each side of the platform, one for each leg. The patient foot position is tracked in real time using a simple marker-based technique. An active marker consisting of a battery-powered infrared light-emitting diode (LED) is placed on the patient heel (sides of the calcaneus bone). The marker size is about 50mm (diameter). The infrared filter of the cameras was initially removed, and a black-film was placed in front of the camera lens to prevent the visible light from being seen by the camera. However, the cameras could see the infrared light of the markers. This system provides both flexibility and safety for the patient, and high robustness to external disturbances, i.e. lighting changes, cluttered environment etc. The cameras are providing high-resolution images at the frequency of 30 Hz, which is adequate for the timing requirements of the control of the treadmill vertical motion during normal walking. Therefore, the position of the patient's foot on the sagittal plane is given in real-time, by using this image-based feedback system. The setup with the cameras and the markers is depicted in Fig. 1, while the processing of the acquired images and the marker detection was discussed in [20].

In [20] we presented the closed-loop control of the MIT-Skywalker for providing the necessary swing clearance for the patient's foot. The concept of the device was illustrated and proved with real-time experiments with unimpaired subjects. However, during the stance phase, the treadmills were just kept at horizontal level, so the interaction with the patient's foot was not controlled. Moreover, the initial conditions of the leg at toe-off were not controlled as well. The control of the initial conditions (position, velocity) of the foot at toe-off could play significant role in increasing the impaired leg range of motion, while the control of interaction of the treadmill with the leg can provide valuable afferent feedback to the patient, which is important for gait recovery. In this paper we control the interaction of the treadmill by introducing a potential field at the foot workspace.

B. Potential field at the foot workspace

A potential field that is applied at the foot should drive it to a desired (or preferred) trajectory by applying forces to the subject's foot, when it is in contact with the treadmill. In order to construct the potential field, we need first to define the desired trajectory for the foot. At this point, it must be noted that the saddle-like seat of the Skywalker allows only passive motion of the pelvis. However the vertical motion of the pelvis during normal walking can be more than $\pm 5cm$. Therefore, if natural gait kinematics are also required, the vertical motion of the pelvis should be accommodated by corresponding motion of the walking surface. In other words, in order to resemble natural gait kinematics, the vertical motion of the pelvis is transferred to the equivalent vertical motion of the walking surface.

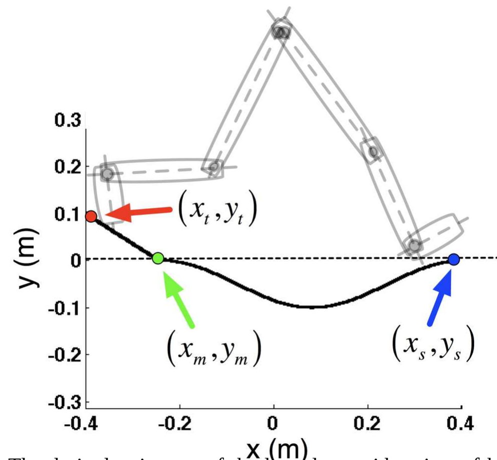

Fig. 3. The desired trajectory of the leg, along with points of heel-strike, mid-stance and toe-off used for the design of the trajectory.

Let heel-strike, mid-stance, and toe-off be the three *phases* that we define as parts of the total stance phase of the gait, i.e. from the time of initial contact of the foot with the treadmill (after the swing phase), until the instance the foot leaves the treadmill (entering the swing phase). At heel-strike the foot should land on the treadmill, when the latter is at level height. Then, the vertical motion of the pelvis is transferred to the walking surface. This motion has an approximately sinusoidal profile. Finally, during the toe-off phase, the desired foot height is linearly increased with respect to the level height. This makes the potential field exert force to the foot upwards, pushing the leg up so to supply it with additional energy to initiate the swing phase.

Based on the vertical motion of the pelvis during natural gait, which approximates one period of a sinusoidal profile for the stance phase, the desired foot trajectory at the vertical axis for the phases of heel-strike and mid-stance is defined by:

$$y_1(x) = A_1 \sin\left(\omega\left(x - x_m\right) + \varphi_1\right) \tag{1}$$

where $x \in [x_m, x_s]$ represents the horizontal axis of motion ranging from a point x_m prior to the toe-off point, to the heel-strike point x_s, A_1 is the magnitude of the sinusoidal profile, $\omega = \frac{2\pi}{(x_s - x_m)}$, and φ_1 is the initial phase of the sinusoidal profile. All the parameters are computed or tuned to match natural profiles for a gait cycle lasting approximately 1 sec, which is typical for normal walking speeds.

The desired trajectory of the foot at toe-off is defined as linearly increasing in height above the level height of the treadmill, and is given by:

$$y_2(x) = \left(\frac{y_t - y_m}{x_t - x_m}\right) x + \left(y_m - \frac{y_t - y_m}{x_t - x_m} x_m\right) \tag{2}$$

where (x_m, y_m) is the point prior to toe-off, and (x_t, y_t) is the toe-off point. All the points, along with the desired leg trajectory profile are shown in Fig. 3.

Regarding the potential field, we opted to define it in such a way that the exerted force profile will be smooth enough and will gradually pushes the leg to the desired trajectory, avoiding high forces close to the desired trajectory. It was finally selected that the force profile for each point in the foot workspace will be given by a 3^{rd} order polynomial function of the vertical deviation from the desired trajectory. Moreover, for each of the phases (heel-strike, mid-stance and toe-off) separate boundary conditions will be defined for exerting different amount of force based on the gait phase. Finally, in between the three different profiles for each phase, two transition phases were also introduced to make the change of the applied forces smoother. For a given position x of the foot along the horizontal axis, the potential function is defined by:

$$V(\Delta y) = a_0 \Delta y + \frac{1}{2} a_1 \Delta y^2 + \frac{1}{3} a_2 \Delta y^3 + \frac{1}{4} a_3 \Delta y^4 \quad (3)$$

where $\Delta y = y - y_0$ is the distance of the foot to the desired trajectory along the y axis. The parameters a_0, a_1, a_2, and a_3 are dependent on the x of the foot along the horizontal axis, and are given by:

$$\begin{aligned} a_0 = 0, \qquad & a_1 = 0 \\ a_2 = \frac{3F_{\max}}{\Delta y_{\max}^2}, \quad & a_3 = -\frac{2F_{\max}}{\Delta y_{\max}^3} \end{aligned} \quad (4)$$

where $F_{\max}$ is the maximum force applied by the force field along the y axis at a distance $\Delta y_{\max}$ from the desired trajectory along the same axis. These two parameters are dependent on the x coordinate of the foot along the horizontal axis, in a way that will be discussed below. Finally, the force applied at each point (x, y) of the foot trajectory is given by:

$$F(x, y) = \frac{\partial V(\Delta y)}{\partial \Delta y} \quad (5)$$

For distances greater than $\Delta y_{\max}$ the force is kept constant and equal to $F_{\max}$.

As mentioned above, the potential will vary along the phases of the stance phase, essentially consisting of three phases for heel-strike, mid-stance, and toe-off, while two transition phases are introduced between them. Therefore for each of the three phases, different values for the parameters $\Delta y_{\max}$ and $F_{\max}$ are selected. The latter are listed in Table I. The final force profile generated by the proposed potential field along with the desired trajectory is shown in Fig. 4. As it can be seen, the forces exerted during heel-strike are high enough for small foot deviations, in order to provide the appropriate feedback to the subject. Then the exerted forces are decreasing, while at the toe-off phase, the forces exerted beyond the level height in order to push the leg up to initiate the swing phase.

It must be noted that the ranges of each of the three phases are a priori defined based on the step length for each individual subject, while the length of the heel-strike and toe-off phase are each one defined to be approximately 15% of the total step length. Finally, the length of each of the two transitional phases is approximately 5% of the total step length.

C. Cam control

As previously noted, the vertical motion and force exerted by the treadmill is controlled through the control of the cam. Let $\psi = f(\theta)$ be the function that describes the profile of the cam in polar coordinates, i.e. the radius of the cam for at each rotation angle θ. This function is known by the design of the cam, and is smooth and continuous [20]. Since the follower of the cam shown in Fig. 2a is always in contact with the cam profile, the vertical motion of it is also equal to ψ with respect to the center of the cam. Then, by assuming no friction at the follower[1], the vertical force that it applies to the treadmill is given by:

$$F = \frac{\tau \frac{d\theta}{dt}}{\frac{d\psi}{dt}} \quad (6)$$

where τ is the torque at the cam axis, $\frac{d\theta}{dt}$ is the rotational speed of the cam, and F is vertical force exerted. Therefore, if $F(x, y)$ is the force exerted by the potential field when the foot is at the point (x, y), the commanded torque at the cam is given by:

$$\tau = F(x, y) \left. \frac{df(\theta)}{d\theta} \right|_{\theta_t} \quad (7)$$

where θ_t is the position of the cam at the specific time instance. It must be noted that the position of the cam is controlled independently and it is not related to the desired trajectory. In other words, since the proposed controller determines the interaction forces between the treadmill and the subject's leg, the desired trajectory is not kinematically

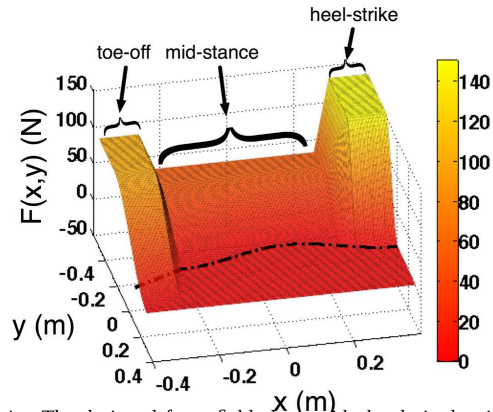

Fig. 4. The designed force field along with the desired trajectory.

TABLE I
PROPERTIES OF THE POTENTIAL FIELD

Phase	Parameter	Value
1- Heel-strike	$F_{\max}$ (N)	150
	$\Delta y_{\max}$ (m)	0.15
2- Mid-stance	$F_{\max}$ (N)	30
	$\Delta y_{\max}$ (m)	0.15
3- Toe-off	$F_{\max}$ (N)	80
	$\Delta y_{\max}$ (m)	0.25

[1]The cam follower has no sideways guides as often used in such systems. Instead, it is free to follow the cam-imposed trajectory, and therefore the friction force is negligible.

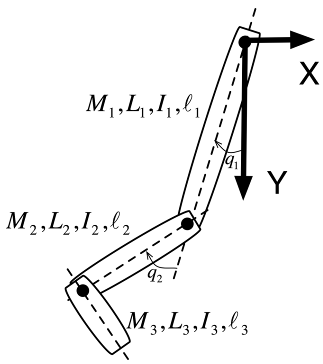

Fig. 5. The simplified dynamic model of the leg used for simulations. The leg is modeled as a 2 degrees-of-freedom mechanism. Each limb segment (thigh, shank, foot) has individual dynamic parameters: mass (M_i), total length (L_i), inertia with respect to center of mass (I_i), distance from joint to center of mass (ℓ_i), $i = 1, 2, 3$.

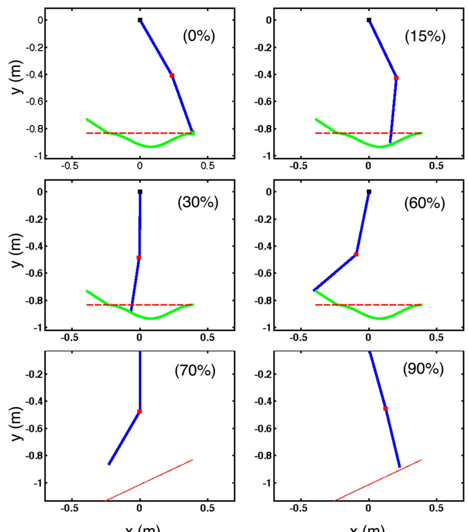

Fig. 6. Simulation results with the potential field acting on the leg during a full gait cycle. Snapshots of heel-strike (0%), midstance (15-30%), toe-off (60%), mid-swing (70%), and late-swing (90%) are shown. The potential field was active only during the stance phase (0%-60%) of the simulation. After that, the treadmill has dropped and the leg swings forward (60%-100%).

imposed by the control of the cam position. Instead, for a given cam position, the cam torque is controlled in order to exert the necessary vertical force, generated by the aforementioned potential field. At this point it must also be noted that since there is no tight connection between the subject's foot and the treadmill, the potential field can not exert forces pulling downwards the leg. This is the reason why the force field is zero above the desired trajectory, as depicted in Fig. 4.

III. Simulation Results

The proposed potential field-based controller was tested in a simulation environment of the MIT-Skywalker. A simplified dynamic model of the lower limb was used, as shown in Fig. 5. The motion of the leg at the sagittal plane is only considered. Moreover, the ankle joint is considered fixed and at neural position. This is a practice that was followed in the experiments with unimpaired subjects too, in order to simulate the prevention of drop-foot and allow

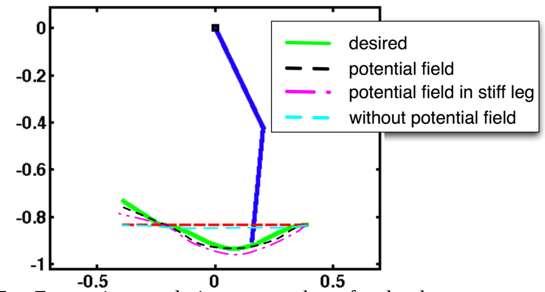

Fig. 7. Foot trajectory during stance phase for the three cases examined.

the treadmill to provide the necessary swing clearance [20]. The parameters of the leg dynamics used in the simulation are reported in Table II.

Initially, the leg is considered completely passive [case 1], i.e. no actuation at the hip and knee joints is modeled. The potential field pushes the leg close to the desired trajectory, not imposing however strict kinematic control, as shown in Fig. 6. Moreover, at toe-off, the proposed controller pushes the foot upwards, allowing for increased hip extension and knee flexion, that increases the total energy of the leg, and facilitates the initial conditions for a proper swing phase.

Muscle tone or high knee stiffness are phenomena usually encountered at stroke patients. In order to test the proposed controller performance in such cases, a stiffness element at both the knee and hip joint was included in the dynamic model of the leg [case 2]. This element was essentially a torsion spring at both joint, while the simulated torque coming from this was defined as linearly varying with respect to the angle of the joint relative to the neural position, i.e.

$$\begin{aligned} \tau_h &= -k_h q_1 \\ \tau_k &= -k_k q_2 \end{aligned} \tag{8}$$

where τ_h, τ_k are the exerted torques at the hip and knee joint respectively, q_1 and q_2 are the hip and knee joint angles respectively, while k_h and k_k are stiffness parameters reported in Table II. Moreover, the behavior of the foot in the case where no external forces were exerted by the potential field [case 3] was also investigated. For such case, the treadmill implements a pure kinematic control of the foot, and at toe-off, it just drops to allow for swing clearance. Fig. 7 depicts the foot trajectory at the three simulated cases. As it can be seen, the potential field-based controller could drive the leg close to the desired trajectory, even in the case when the leg joints are stiff.

In order to quantify how the proposed potential field aids the leg swing phase by providing energy at the system just before toe-off, the energy of the foot for a full gait cycle was computed, using the following equation:

$$E = \frac{1}{2}M_1 v_1^2 + \frac{1}{2}M_2 v_2^2 + \frac{1}{2}I_1 q_1^2 + \frac{1}{2}I_2 q_2^2 + M_1 g y_{c1} + M_2 g y_2 \tag{9}$$

where v_1, v_2, y_{c1}, y_{c2} are the velocities and heights with respect to the horizontal axis of the center of mass of the thigh and shank respectively, and M_1, M_2, I_1, I_2 are masses and moments of inertia of thigh and shank respectively. The resulted values of energy just before toe-off for the three

TABLE II
SIMULATION PARAMETERS

Leg Dynamics					
$M_1\ (Kg)$	8.00	$M_2\ (Kg)$	3.72	$M_3\ (Kg)$	1.16
$L_1\ (m)$	0.47	$L_2\ (m)$	0.45	$L_3\ (m)$	0.27
$I_1\ (Kgm^2)$	0.32	$I_2\ (Kgm^2)$	0.14	$I_3\ (Kgm^2)$	0.06
$\ell_1\ (m)$	0.20	$\ell_2\ (m)$	0.19	$\ell_3\ (m)$	0.11

Hip & Knee Stiffness Parameters			
$k_h\ \left({}^{Nm}/_{rad}\right)$	25	$k_k\ \left({}^{Nm}/_{rad}\right)$	20

cases were 8.22, 7.78, 2.3 (J) respectively. Therefore, the energy of the system just before toe-off is increased in the case where the proposed controller was used, even in the case of the stiff leg joints. For the case of non-active potential field, the leg has only potential energy when it initiates the swing phase. This could limit the range of motion for the leg, especially in the case where the paretic leg has increased joint stiffness, due to for example muscle tone.

IV. CONCLUSIONS AND DISCUSSION

As robot-assisted gait therapy is increasingly gaining acceptance at rehabilitation centers, the MIT-Skywalker may prove to be the most effective and low-cost gait rehabilitation device. The fast don and doff alongside its dynamic principle and ecological intervention may place this novel device apart from the existing kinematically-based rehabilitation devices. Finally, the proposed potential field-based control scheme may prove very efficient in delivering both the desired sensory input to the patient and for increasing the leg's range of motion, which would aid in his/her regaining lower limb control and walking ability.

Finally, the system affords future integration with our Anklebot [7] and pelvis robot (Elvis-the-Pelvis [21]), thereby allowing entire lower body training. One has to apply the appropriate caveats to this manuscript: until we can collect clinical data with patients and evaluate who might benefit and what are the potential outcomes, we cannot evaluate the device actual impact in the field. To that end, we will be deploying the MIT-Skywalker to the VA Baltimore shortly.

ACKNOWLEDGMENTS

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