Modeling and Simulation of a Lower-body Wearable Exoskeleton Using Robotics' Techniques

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***Abstract*—The article presents design and preliminary simulation results for a lower-body exoskeleton system. The device has 5DoF per leg and a common hip rotation degree between both legs. The mathematical model of the combined human-exoskeleton system is developed using the Robotics Toolbox in Matlab. Simulations are carried out to estimate joint torque and power profiles for straight-line on-ground walking. The results indicate the validity of the developed model and show promise for the design of links and actuation systems in future.**

***Index Terms*—Legged locomotion, wearable robots, exoskeletons**

I. Introduction

Spinal cord injuries (SCI) are a common phenomenon worldwide due to several reasons including road accidents, falls and violence. These injuries usually result in partial or full paralysis of certain body parts, including the portion of the body below waistline; a condition known as paraplegia. The most significant impairment of paraplegia is the loss of mobility (paralysis). Another major cause of lower-body paralysis is the epidemic of Poliomyelitis (or infantile paralysis). There is an abundance of such patients in low-income countries in Asia and Africa. According to the World Health Organization estimate, there are at least over a million polio survivors throughout the world while traumatic SCI affects up to 500000 people every year [1].

Such patients are mostly restricted to wheelchairs and are fitted with fixed braces and splints known as Orthoses to allow limited mobility. Orthoses are externally attached structures, which are inexpensive and have been around since long. However, the biggest limitation of orthoses is the absence of external power. This is because the orthoses are fundamentally designed to provide stability rather than ambulation. On the other hand, powered exoskeletons seem to provide the missing component in orthoses for paralyzed patients, that is, extra power. The structure is equipped with sensors and motors to predict human intention and provide support to the limb accordingly. The purpose of our research is to propose an inexpensive powered exoskeleton device for patients in low-income countries.

Exoskeletons have a long history of development. The first exoskeleton appeared in 1969 in Mihajlo Pupin Institute Belgrad [2]. Similarly, General Electric (GE) developed Hardiman from 1965 to 1971. This was a full-body exoskeleton. Both these machines consisted of massive structures (in the range of a few hundred kilograms) and were either developed for medical purposes or to enhance capabilities of otherwise able-bodied humans such as soldiers. These systems are considered the predecessors of the modern humanoid robots. After some pause, exoskeleton started appearing again in the 21st century building on the advances made in the field of humanoid robots. These modern exoskeletons are mostly aimed at medical use which is also the objective of our research.

One of the most advanced exoskeleton to date is HAL (short for hybrid assistive limb) developed at Tsukuba University Japan [3]. It works by detecting the bio-electric signals on the surface of the skin and provides adequate power to joint motors. It has two versions, HAL3 targeting only the leg function and HAL5 targeting full body. Another advanced and commercially available exoskeleton is ReWalk [4]. It is intended to provide power to hip and knee for patients with spinal cord injury. However, it is supplemented with crutches to support human weight. Other powered exoskeletons include BLEEX [5], LOPES [6] and Ekso. The cost of these devices is prohibitively high, ranging from US$30,000 to 100,000.

Apart from the cost, since most of the above-mentioned systems are intended for commercial use, their design parameters and characteristics are not well communicated. During the course of development, there are several design and control decisions that may affect the performance of the device such as the type of actuation, the size and location of actuators, the type of control laws etc. These decisions greatly affect the overall energy consumption of the device and are hence important to take into account earlier. Based on these observations, before we go to the full development of our system, we intend to harness the power of mathematical modeling to make informed decisions on the above-mentioned aspects, especially power system requirements and control laws.

In this article, we present the basic design and preliminary simulation results for our combined human-

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exoskeleton model. The objective of these simulations is to get an insight into the torque profiles at different joints and the power requirements. Later on, the model shall also be exploited to test different control schemes. To our knowledge, there is no such model available for this purpose.

The article is organized as follows: Section II describes the design approach and the mathematical modeling of the system. Section III presents simulation results while Section IV concludes the article.

II. Methods

In order to design an exoskeleton device, study of human anatomy is essential. This is because the system has to work in parallel with the human body and any mis-alignment may lead to discomfort. Human body is divided into 3 planes namely sagittal, coronal and transverse. Straight-line walking mainly takes place in the sagittal plane with small motions in the other two planes.

Human joints typically provide complex interaction between adjoining limbs. However, they can be biomechanically simplified in terms of the number of free rotations they allow (a.k.a Degrees of Freedom or DoF). To this end, a human leg can be considered as a 7-DoF system. Most lower-body exoskeleton devices incorporate a similar joint architecture. Below we present the design of our device with its comparison to the biological joints:

- Hip Joint: A human hip joint allows 3 rotational degrees of freedom namely flexion/extension, abduction/ adduction and internal/external rotations through a single ball and-socket joint. However, for simplicity, we decouple the hip joint into 3 single-DoF joints whose center of rotation passes through the human hip joint. However, a common axis is chosen for internal/external rotations of both legs in the middle. This choice is made to avoid unnecessary complications regarding singularities as discussed in [5].
  Moreover, the relative distance between the hip joint axes is made variable so that it can fit to people with varying waist sizes. This is achieved using a set of externally shafts with sliding collars.
- Knee and ankle joints: Human knee is a condyloid joint which allows motion in various directions except for the axial rotation. However, the principal rotation of knee joint in walking is the flexion and extension in the sagittal plane. Therefore it is modeled as a hinge joint in our model. The ankle joint is modeled as a decoupled 2-DoF joint allowing rotation in sagittal and coronal planes. Just like the hip assembly, the lengths of thigh and lower-leg links are also made variable. The link is constructed in 3 parts with internal and external threading which fit together at different lengths.

The 3D CAD model of the system is developed in SolidWorks and at present are not actuated (Fig. 1).

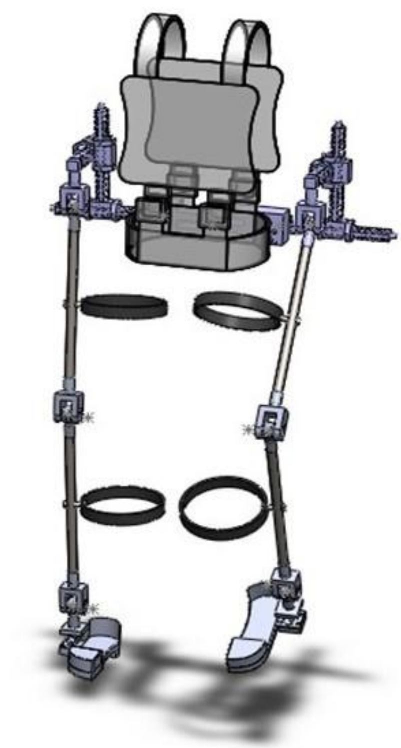

Figure 1.The CAD model of the complete system

A. Fabrication of Prototype

A life-size prototype of the system, with a simple limb structure, is developed to test the wearability and maneuverability of the device (c.f. Fig. 2). The leg links are machined with solid aluminum circular bars. The joints are sand cast. The links are named progressively, starting from the right foot and moving up to the hip and finishing at the other foot. The total mass of the exoskeleton system is less than 11kg, excluding future actuation and other payloads, which shall be designed after an insight from our mathematical model. The system shall be used in conjunction with a mobility aid such as crutches and/or a body weight support system.

B. Mathematical Model

In order to generate the mathematical model, the exoskeleton is considered as a spatial system of articulated bodies. A systematic parameterization is necessary to describe its structure. The most popular notation in this regard is the one proposed by Denavit & Hartenberg in 1955. It consists of attaching coordinate frames to each link of the system and writing 4 (instead of 6) standard parameters to describe their position and orientation with respect to neighboring links. It can be shown that for N+1 links connected by N single DoF (revolute) joints, these $4 \times N$ parameters uniquely define the position and orientation of each link of the system in 3D space.

Figure 2. The life sized prototype wore by an able-bodied user

This method has long been used to describe industrial robotic manipulators. However, the method can also be extended to anthropomorphic structures such as humanoid robots and exoskeletons.

When applying this method to exoskeletons, the first difficulty arises while choosing the base frame. Since the system does not have a fixed base, unlike industrial manipulators, the choice of the base link is not a straight-forward one. In the field of humanoid robots, 2 distinct approaches are taken.

The first approach takes the base at the hip level where the legs and trunk sections are attached making a treelike structure. This system is considered kinematically free without inherent contact dynamics with the environment [7], [8]. The second approach considers the system as a rooted kinematic chain with the base taken at the foot on ground [9]. We choose this latter modeling approach as it results in a smaller number of configuration parameters at any time instant. This way, the complete exoskeleton can be considered as being rotating around the stance foot (link 0). The foot in the air is considered at the terminal link (link 11) or the end-effector. Fig. 3 shows the frame attachment to the exoskeleton links considering right foot on the ground. One limitation of this approach is that the system is always considered with one foot in contact with the ground. The double-support of human walking results in a closed-loop structure and requires separate treatment.

These parameters are used to generate a geometric model of the system in Matlab using the Robotics Toolbox [10]. The toolbox is further used to generate kinematic and dynamic models for joint torque estimation.

C. Dynamic Model and Human Limbs Integration

Once the kinematic model of the system is available, its dynamic model can be derived given the inertial parameters of all the links and joint motion profiles. The inertial parameters include link masses, moment of inertia values and the center of mass (CoM) locations in the respective link frames. For the exoskeleton segments, these parameters are easily calculated from the CAD model and the physical prototype.

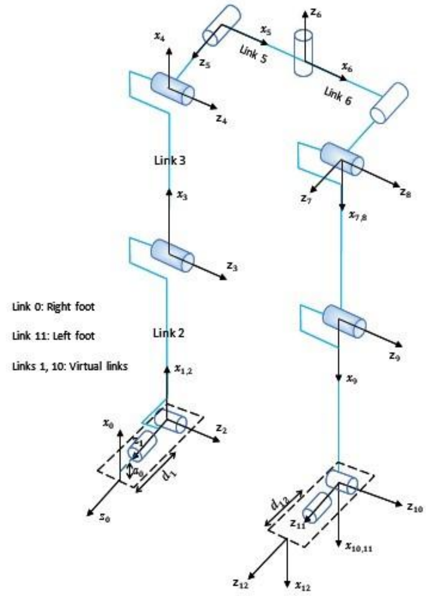

Figure 3. Frame attachment of to the system using the modified Denavit-Hartenberg notation

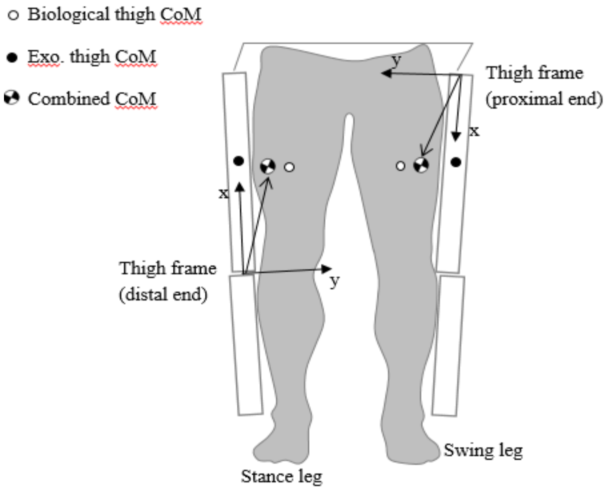

Figure 4. The calculation of inertial parameters according to their respective frames for swing and stance legs

However, the exoskeleton has to work parallel to the human leg with all the power provided by the external actuators. Hence for meaningful computations, the model has to take into account the average human limb characteristics. For this, we turn to the regressions found in the biomechanics literature mainly [11] and [12] for average male parameters (male, 73kg, 1.74m). The limb masses are simply added to that of the corresponding exoskeleton links. Since our exoskeleton does not have an upper-body segment, the mass of the human upper body (HAT) is added to the hip links (5 and 6, Figure 3). Combined CoM location values for the biological and exoskeleton limbs are calculated by adding the products of segment masses and individual CoM locations, divided by the cumulative mass of both segments. The combined moment of inertia values are calculated at the proximal (for swing leg) and distal (for stance leg) ends of the links in the exoskeleton frame (c.f. Fig. 4). The process is repeated along all 3 axes.

In order to simulate the system, the joint motion profiles are taken from the experimental biomechanics literature for normal flat-ground walking and averaged across subjects [13]. Motion of only 2 joints (hip and knee) in the sagittal plane is considered. The foot is considered flat with respect to the ground at all times. In order to reproduce the joint trajectories, the human trajectory is divided into 2 sub-phases (stance and swing) and then separately approximated using the 4th degree polynomial functions. These functions are then differentiated to get angular velocity and acceleration profiles needed for dynamic calculations. The data is given for a walking speed of 1.2m/s. However, for our simulations, the speed is reduced by half. The simulations are performed in an open-loop fashion to calculate the joint torques through inverse dynamics. The total mass of the system (human + exoskeleton) is around 84kg.

III. Results

Fig. 5 shows the time elapsed images of the whole system executing the walking motion. A snapshot is taken at different times, indicating the mid-stance and foot contact instants. A total of 3 steps are taken in 2.5s, starting from the left foot. The built-in functions of the robotics toolbox are modified to improve visualization and to include custom annotations.

Fig. 6 shows the components of the ground reaction for the right leg. The motion starts from an upright posture with both feet on the ground. The shapes of the curves are similar to what is observed in actual level ground human walking [11]. The horizontal component (blue) gradually changes sign around mid-stance while the vertical component (black) modulates around the body weight due to small knee flexion during stance. Note that the first stance phase is smaller as the system is considered to start from the upright posture, unlike subsequent steps which start from behind. There are clear discontinuities at the step transition instants due to the absence of double support phase, which is responsible for smooth transition between the steps. All in all, the force curves validate our modeling approach alongwith its inherent limitations.

Subsequent figures show our preliminary results for power and torque calculations. Fig. 7 shows the time profile of joint powers (per kg) for hip and knee joints of right leg for all 3 steps. Very high power consumption is observed at the joints, especially at the end of a phase where the feet change roles. We attribute this peak to the discontinuity of motion at this point owing to the absence of a double support phase. It is known that this phase allows for the smooth transition of center of mass velocity of the system from downward to upwards [14]. The existing framework of Robotics Toolbox does not allow the incorporation of a double-support phase as the system departs from the serial architecture. We shall seek to implement this feature to incorporate the smooth transition of velocities and accelerations from one step to another.

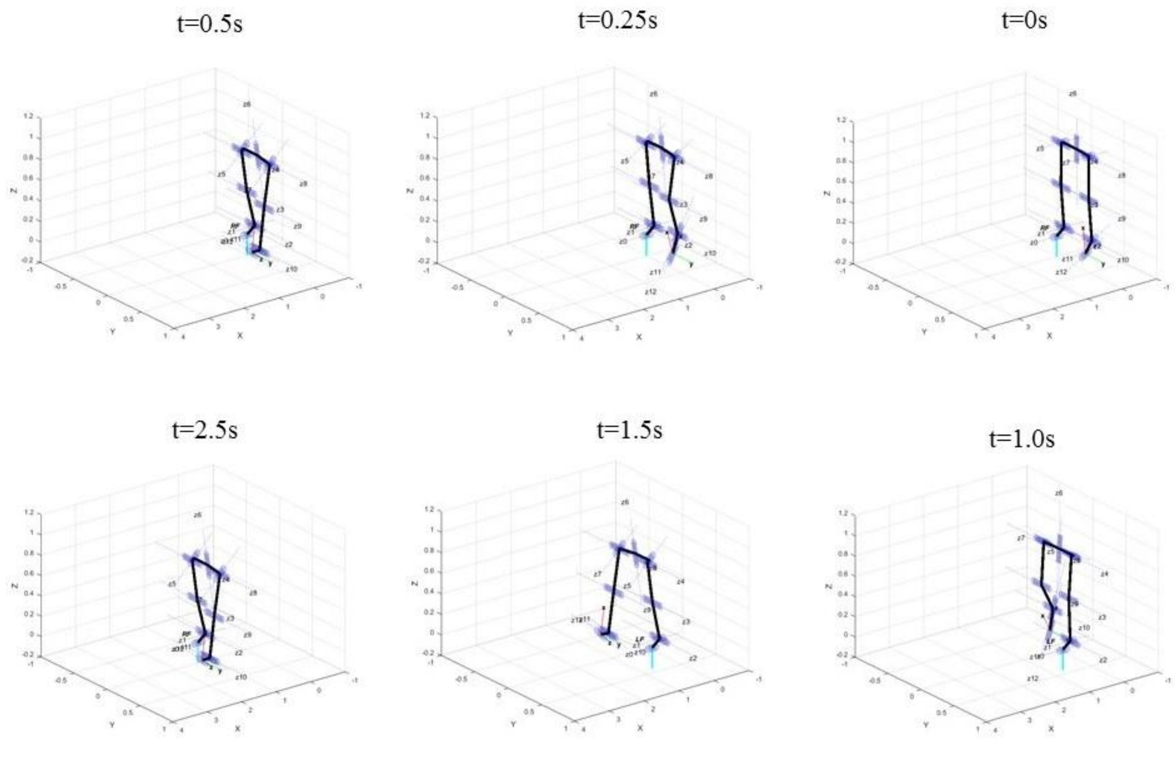

Figure 5. The time-elapsed images of the system undertaking the walking motion generated using the Robotics Toolbox in Matlab

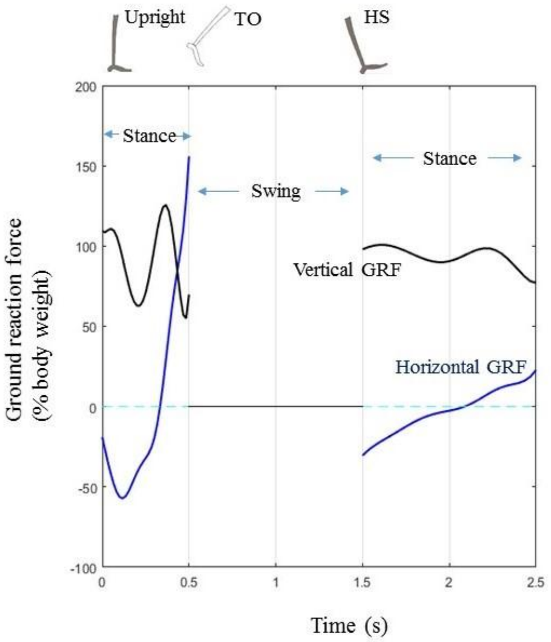

Figure 6. Horizontal (blue) and vertical (black) components of ground reaction force acting on the right foot

Finally, Fig. 8 shows the torque-velocity relationship for the hip and knee joints of the right leg. High instantaneous torques are observed at the relatively low speeds during stance, especially at the start and end of the phase. Such torques would lead to the selection of bigger actuators and larger batteries for autonomous operation. However, we believe that these peak values can be mitigated by designing efficient controllers that exploit natural human dynamics and by introducing the double support phase between the steps. In this regard, the framework presented in this article shall be valuable in evaluating the performance of different controllers.

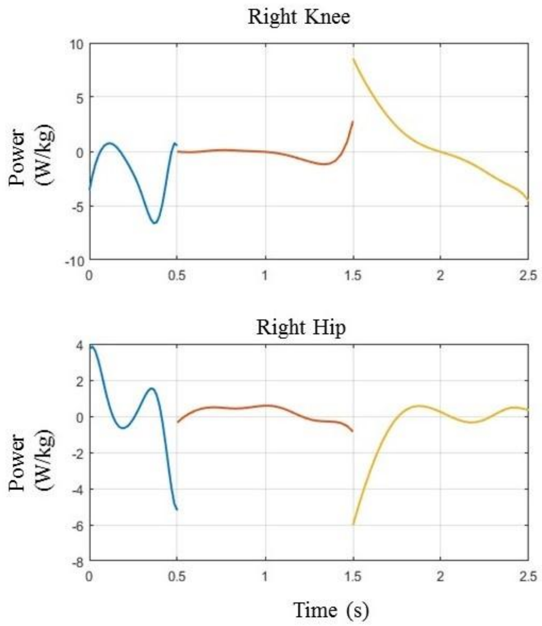

Figure 7.Instantaneous power (W/kg) at knee and hip joints of the right leg

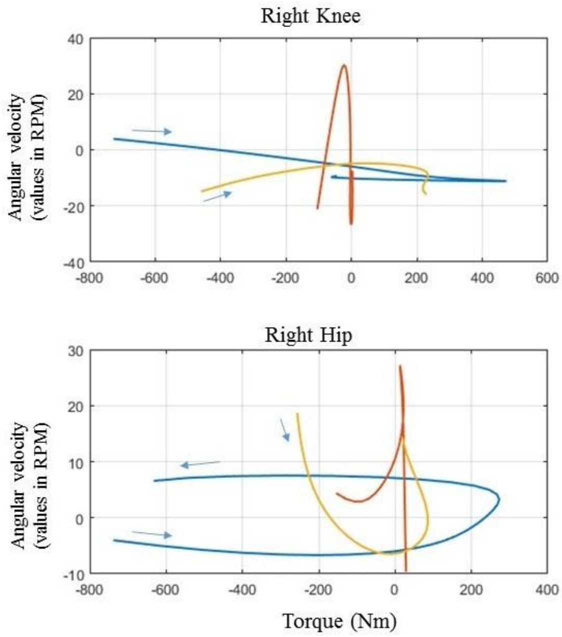

Figure 8.The torque-velocity curves for knee and hip joints of the right leg during all 3 steps. Peak torque values coincide with the start and end of stance phases

IV. Conclusions

Though the results show the usefulness of our model, they also highlight some shortcomings. The inability of the model to represent double-support phase for smooth redirection of velocities results in large discontinuities at impacts. In future, we shall improve this model by including a double support phase of waking, friction and heat losses in motors as well as in transmission. Moreover, we shall test different controllers and observe their effects on the power/energy requirements.

This will allow the selection of appropriate actuators to power our device.

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