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Rendering potential wearable robot designs with the LOPES gait trainer

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***Abstract*— In recent years, wearable robots (WRs) for rehabilitation, personal assistance, or human augmentation are gaining increasing interest. To make these devices more energy efficient, radical changes to the mechanical structure of the device are being considered. However, it remains very difficult to predict how people will respond to, and interact with, WRs that differ in terms of mechanical design. Users may adjust their gait pattern in response to the mechanical restrictions or properties of the device. The goal of this pilot study is to show the feasibility of rendering the mechanical properties of different potential WR designs using the robotic gait training device LOPES. This paper describes a new method that selectively cancels the dynamics of LOPES itself and adds the dynamics of the rendered WR using two parallel inverse models. Adaptive frequency oscillators were used to get estimates of the joint position, velocity, and acceleration. Using the inverse models, different WR designs can be evaluated, eliminating the need to build several prototypes. As a proof of principle, we simulated the effect of a very simple WR that consisted of a mass attached to the ankles. Preliminary results show that we are partially able to cancel the dynamics of LOPES. Additionally, the simulation of the mass showed an increase in muscle activity but not in the same level as during the control, where subjects actually carried the mass. In conclusion, the results in this paper suggest that LOPES can be used to render different WRs. In addition, it is very likely that the results can be further optimized when more effort is put in retrieving proper estimations for the velocity and acceleration, which are required for the inverse models.**

Keywords: gait, rehabilitation robots, wearable robots, adaptive frequency oscillators.

I. Introduction

To assist physically disabled, injured, and/or elderly persons, a wide variety of supportive devices are being developed. These devices can consist of exoskeletons, prostheses, or other wearable mechatronic devices and can be used for several applications, such as rehabilitation, personal assistance, human augmentation, and more.

Robotic applications for training and assistance have rapidly evolved during the last decade. The first generation of robotic devices was mainly directed at providing gait training in a controlled environment. They perform repetitive tasks and are often used in combination with a treadmill. These traditional robotic gait trainers are now expanding to mobile systems, that can be used outside the clinic and that can assist or augment patients during several activities of daily living. These mobile devices are referred to as Wearable Robots (WRs). A WR is classically defined as a mechatronic system designed around the shape and function of the human body, with segments and joints corresponding to those of the person it is externally coupled with [1]. These WRs are expected to interact and collaborate with the user in an intelligent manner.

There are several challenges that must be faced to successfully introduce WRs as supportive or augmenting mobile devices. The main challenge lies in reducing the metabolic energy consumption of the user, while simultaneously minimizing the energy requirements of the actuators. The problems of energy requirements are being tackled on several fronts. On the one hand, more efficient actuators and more powerful batteries are being developed. On the other hand, changes are being made to the structural design of the WR. These changes can consist of making the WR more lightweight, or adding springs to the joints that store and release energy throughout the gait cycle.

Thus, for the new generation of WRs, it is important that the mechanical design is energetically optimized. This can lead to designs that do not follow the classical definition of WRs, where the human and mechanical joints coincide. However, it is very difficult to predict how people will react to, and interact with, these new types of WRs. Users may adjust their gait patterns in response to the mechanical restrictions or properties of the device. In this paper we propose a new method to simulate different WR designs using the existing gait trainer LOPES (Lower Extremity Powered ExoSkeleton).

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The goal of this pilot study is to show the feasibility of rendering the mechanical properties of different potential WR designs with LOPES. We developed a method that selectively cancels the dynamics of LOPES itself and adds the dynamics of the rendered WR. In this manner, different WR designs can be evaluated, eliminating the need to build several prototypes.

II. EXPERIMENTAL SETUP AND METHODOLOGY

A. Subjects

Three healthy subjects (two males and one female, age: 25.7 ± 1.5 years, height: 1.81 ± 0.05 m, weight: 79.3 ± 11.7 kg) participated in this experiment. All subjects gave written informed consent to participate.

B. Experimental apparatus and recordings

To render different mechanical WR designs, LOPES was used. LOPES (see Fig. 1) is a treadmill-based lower-limb exoskeleton type robotic gait trainer. LOPES is impedance-controlled, has eight actuated degrees of freedom (DoF) (flexion/extension at the hip and knee, abduction/adduction for both legs and horizontal pelvis translations), and is initially designed to provide supported treadmill training for stroke patients. It is torque-controlled by means of series elastic actuation (SEA) [2]. Every DoF of LOPES is fitted with potentiometers that record the kinematics. Matlab xPC (Mathworks, Natick, Mass., USA) is used to control the applied torques by the exoskeleton joints at 1000Hz.

The treadmill is equipped with four force sensors capable of detecting the centre of pressure (CoP) during gait. CoP and kinematics are used to detect the different gait phases. All signals are sampled at 100Hz and stored for later processing.

Additionally the interface between the subject's legs and the exoskeleton legs is sensorized using three force sensors. Interface forces and torques are measured with six DoF force sensors (ATI-Mini45-SI-580-20, ATI Industrial Automation, Apex, N.C., USA), which are positioned between the connection cuffs and the exoskeleton frame (see Fig. 2). The cuffs (Hocoma, Volketswil, Switzerland) used in LOPES are made of a rigid carbon fiber shell with Velcro straps and affix the subject's legs to the robot. One cuff connects to the upper leg and two cuffs connect to the lower leg of the subject. Only the interface of the left leg is fitted with force sensors. The analog signals are sampled at 1000 Hz using a data acquisition system (NI usb-6259, National Instruments, Austin, Texas, USA) and sent to the computer, where the data is stored for further processing.

As a predictor of the metabolic energy consumption the subject's heart rate was used. Heart rate was recorded with a wireless POLAR RS400 heart rate monitor (Polar, Oulu, Finland).

Muscle-activation patterns were recorded by bipolar surface electromyography (EMG) from the rectus femoris, vastus lateralis, semitendinosis, biceps femoris, tibialis anterior, gastrocnemius, and the gluteus maximus. All recordings were performed on the left leg only.

Figure 1. LOPES is an bilateral exoskeleton with the actuators detached from the exoskeleton. Series elastic joints are actuated via Bowden cables. LOPES has eight degrees of freedom that are impedance controlled.

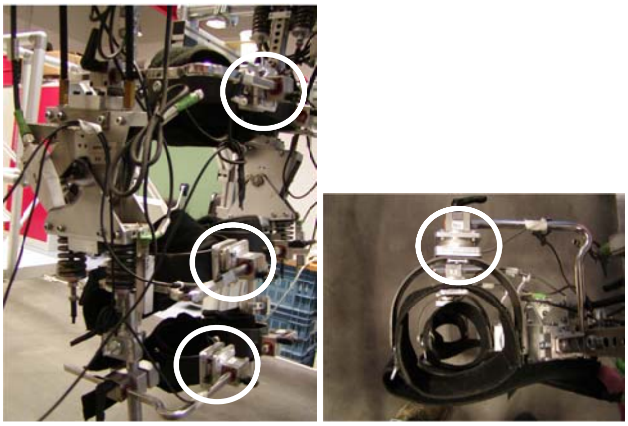

Figure 2. Six DoF force sensors in combination with carbon shells and Velcro straps that attach the upper and lower leg of the subject to the exoskeleton. Three force sensors are positioned between the connection cuffs and the exoskeleton frame (white circles).

Skin preparation and the placement of the disc-shaped solid-gel Ag/AgCl-electrodes in a bipolar configuration were performed according to Seniam guidelines. For the EMG recordings, a compact measurement apparatus (type Porti 16-5, supplier: TMS International, Enschede, The Netherlands) was used. The analog signals were sampled at 1024 Hz and sent from the portable unit via fiber optics to the computer, where the data is stored for further processing. A sync signal was used to synchronize the EMG, LOPES and force sensor data.

C. Controller design

Fig. 3 shows the control strategy used to render the mechanical dynamics of a specific WR design on the LOPES. The first step in simulating the effect of any arbitrary WR design with LOPES is to cancel the dynamics of LOPES itself. This is done using a parameterized inverse model of the LOPES. In this model, the mechanical behavior of the LOPES setup is represented by two double pendulums. Each double pendulum represents one leg of LOPES, consisting of an upper- and lower leg segment. Each segment of the pendulums has a mass (located at a certain distance from the proximal joint) and inertia. Additionally, each joint has rotational damping that represents the friction in each joint. The parameters corresponding to the different LOPES segments are estimated using multi-input-multi-output (MIMO) system identification [3].

The input of the inverse model consists of the hip and knee angle, angular velocity, and acceleration. Currently, LOPES is not fitted with accelerometers that measure the required signals directly. Deriving the velocity and acceleration purely based on the potentiometer signal will generate noisy derivatives. Filtering is therefore required, but unavoidably introduces delays.

To compensate for the delays caused by the filters, adaptive frequency oscillators (AFOs) are used. These oscillators were developed by Righetti and colleagues [4, 5]. The artificial oscillator was based on an augmented Hopf oscillator. From a sinusoidal input, this dynamic system extracts the instantaneous movement features, namely frequency, amplitude, and offset, while keeping its output phase synchronized with the input. This is possible because the oscillator exploits the a priori knowledge that the movement is periodic. Since the knee angle in particular does not follow a sinusoidal pattern, the adaptive oscillator is coupled to a non-linear filter, learning the trajectory envelope [6]. Preliminary experiments showed that this framework can easily be used to estimate the joints' position in the future. In this way, the delay caused by the filter can be compensated for by the oscillator, effectively creating a zero-phase lag filter.

The derivative of the filtered and phase-shifted potentiometer signals yields the angular velocity and acceleration. The obtained states serve as input for the inverse model of LOPES and result in a set of hip and knee torque commands that are sent to the torque controller of LOPES. Using this configuration, walking in LOPES should feel more transparent, since the mass and inertia of the LOPES segments are now compensated for. Parallel to this model, a second inverse model will be used to calculate the torques that would have been exerted to the human when it would have carried the WR. The resulting torques are again sent to the torque controller of LOPES.

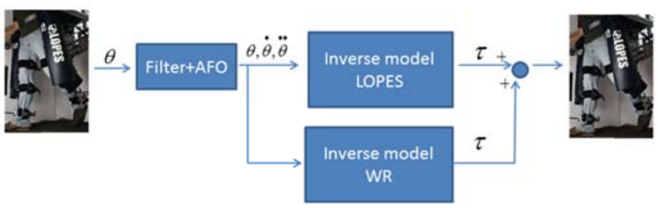

Figure 3. Schematic representation of the proposed method to render the mechanical characteristics of a specific exoskeleton design on the LOPES. The inverse model of LOPES eliminates the LOPES dynamics and the parallel inverse model of the WR is used to render the dynamic behavior of the WR.

D. *Experimental protocol*

Before positioning the subject in the LOPES, different anthropometric measurements were taken to adjust the exoskeleton segments lengths. Additionally, the position of the cuffs was adjusted in two DoFs to align the subject's knee and hip axis with the exoskeleton joints. Next, the subject was positioned into the LOPES and the trunk, thigh, and upper- and lower shank were strapped to the exoskeleton. To let the subject become familiar with the device, every subject walked for five minutes in the LOPES with a constant speed of 3 km/h before testing began.

After this familiarization period, different conditions were tested. Each condition was tested for five minutes with a two-minute resting period in between. Table 1 lists the six different conditions. During the "free walking" condition, the subject walked freely on the treadmill without being strapped to the LOPES. During the "free walking + real WR" condition, the subject walked freely on the treadmill but had the WR attached to the leg. Since this study is more about the proof of principle, and for practical reasons, we chose to simulate a WR that consisted of a point mass at the ankles. This meant that during condition 2, the subject had to walk with two actual weights attached to his ankles. Two SCUBA-diving belt lead weights (3.7 kg each) were used as point masses and fixed to the ankles using Velcro straps. Weights of 3.7 kg were used because previous studies showed that weights of approximately that mass significantly change gait kinematics and EMG patterns [7]. In the "zero-impedance mode," the subject walked in LOPES while no additional torques were exerted. In the "transparent mode," the inverse model of LOPES was used to make LOPES more transparent, and consequently walking more natural. In the "transparent mode + simulated WR", the WR is simulated while the inverse model of LOPES runs in parallel. In the "transparent mode + real WR," the lead weights were used again as a representation of the WR.

All conditions were randomized to minimize the effects of fatigue. During all trials the subjects walked at 3 kph. The total protocol had a duration of two hours (EMG preparation included).

TABLE I. LIST OF TESTED CONDITIONS

Condition number	Definition
1	Free walking
2	Free walking + real WR
3	Zero-impedance mode
4	Transparent mode
5	Transparent mode + simulated WR
6	Transparent mode + real WR

E. Data analysis.

All signal processing was done with custom-written software in Matlab (Natick, Mass., USA). Gait kinematics recorded with LOPES were low-pass filtered with a second-order, zero-lag Butterworth filter with a cutoff frequency of 10 Hz. The measured forces and torques from the three force sensors were resampled to a frequency of 100 Hz and subsequently low-pass filtered with a second-order, zero-lag Butterworth filter (10 Hz). The raw EMG recordings were band-pass filtered (10-400 Hz) with a second-order zero-lag Butterworth filter to remove movement artifacts, full-wave rectified, and low-pass filtered with a low-pass second-order zero-lag Butterworth filter (5 Hz) to smooth the signal. All recorded signals, between the fourth and fifth minute, were broken down into the individual stride cycles, based on the heel-contact event detected by the phase detection algorithm. Next, the different data blocks were normalized as a percentage of the stride cycle and averaged. The average EMG profiles per muscle were transformed into a quantitative measure by integrating the average activity of the stride cycle. The average interaction force was transformed into a quantitative measure by taking the mean value of the absolute force over the stride cycle. Only the force in the sagittal plane and perpendicular to the exoskeleton legs was taken into account. For every condition the mean heart rate and stride time between the fourth and fifth minute was calculated.

III. RESULTS

Fig. 4 shows a typical pattern for the hip and knee angle, recorded by the potentiometers (in red and green respectively) and the reconstructed AFO signal (black dotted line). The figure confirms that the AFO is capable of reconstructing the input signal and cancelling the phase lag created by the filter. The main discrepancy between the true and reconstructed signals occurs in the knee angle signal. The maximum knee angle slightly decreases and a small undershoot at the end of the swing phase can be detected. Both effects can be attributed to the filtering characteristics of the AFO.

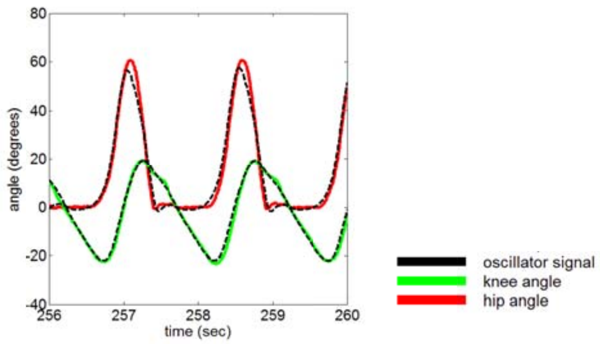

Figure 4. Typical angle patterns for the hip and knee, recorded by the potentiometers (in red and green respectively) and the reconstructed signals by the AFO (black dotted line) (subject 1, condition 3).

To compare walking in the transparent mode with the zero-impedance mode, the EMG levels of both conditions are normalized to the EMG levels during the free-walking condition. Although only three subjects have been measured so far, we do see some trends. Walking in LOPES increases the EMG levels above the level of free walking (dotted line in Fig. 5). In the transparent mode (green) the EMG levels decrease, but do not reach the level of free walking. This suggests that walking in the transparent mode resembles free walking more, compared to the zero-impedance mode, but is not yet as efficient as free walking itself.

This is also confirmed by a slight reduction in interaction force, shown in Fig. 6. This figure shows the interaction force for the different sensorized cuffs during the transparent trials. Here, the interaction forces are normalized to the interaction forces recorded during the zero-impedance trials. On average, there is a decrease in interaction force for all cuffs, indicating that the subject is less hindered by the exoskeleton.

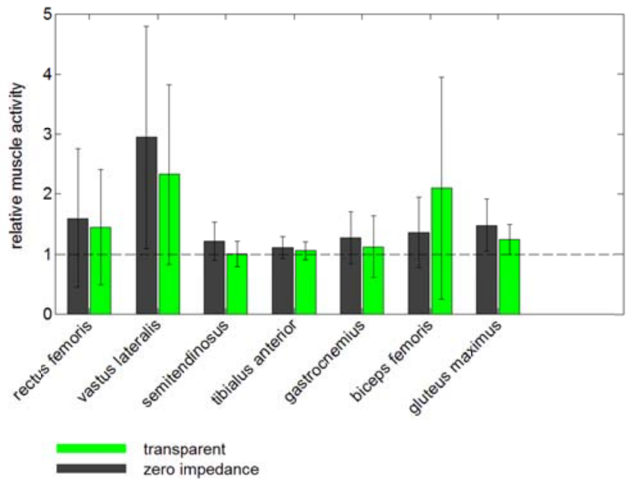

Figure 5. Mean relative EMG levels for the different muscles for the zero-impedance (gray) and transparent (green) mode. EMG levels are normalized to the free-walking trial and averaged across the subjects.

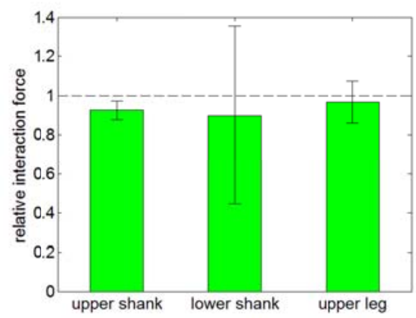

Figure 6. Mean relative interaction force for the different sensorized cuffs for the transparent mode. Interaction forces are normalized to walking in the zero-impedance mode and averaged across the subjects.

To investigate the net effect of the added or simulated weight to the ankle, the EMG levels of the "free walking + real WR" are normalized to the free-walking trials and the "transparent mode + simulated WR" and "transparent mode + real WR" are normalized to the "transparent mode." Fig. 7 shows that the EMG levels increase most when the mass is applied when the subjects are walking freely (cyan bars). It also shows that simulating the mass at the ankle (on top of the transparent mode) increases EMG levels (blue bars), but not to the same level as actually carrying the mass (red bars). For most muscles, the increase in EMG, as a result of the simulated mass, is lower than the increase in EMG due the actual mass.

Fig. 8 shows the stride time and heart rate for the different conditions. Both parameters were normalized to free walking. The figure shows similar trends as observed before. First, the transparent mode decreased the stride time more toward the free walking baseline, compared to the zero-impedance mode. Second, the stride time increases most when the mass is applied when the subjects are walking freely. Similar to the EMG levels, the increase in stride time, as a result of the simulated mass, tends to be lower than the increase in stride time due to the actual mass. The heart rate data also shows that the increase in heart rate due to the actual mass is larger than the increase in heart rate due to the simulated mass. Walking in the zero-impedance or transparent mode does not seem to influence the heart rate much.

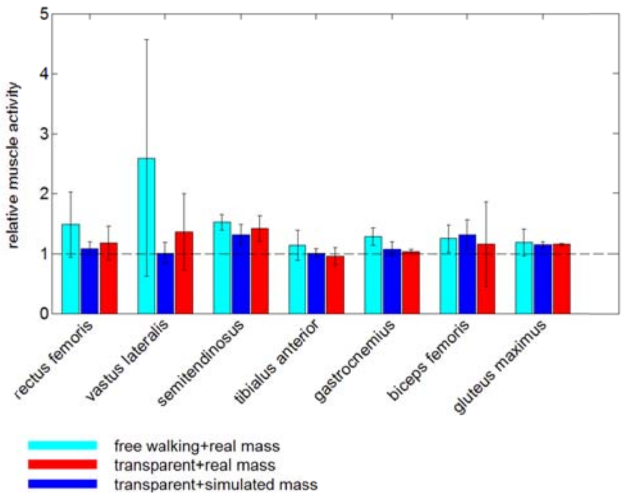

Figure 7. Mean relative EMG levels for the different muscles for the "free walking + real mass" (cyan), "transparent + real mass" (red) and "transparent + simulated mass" (blue). EMG levels are normalized to the transparent or free-walking trial and averaged across the subjects.

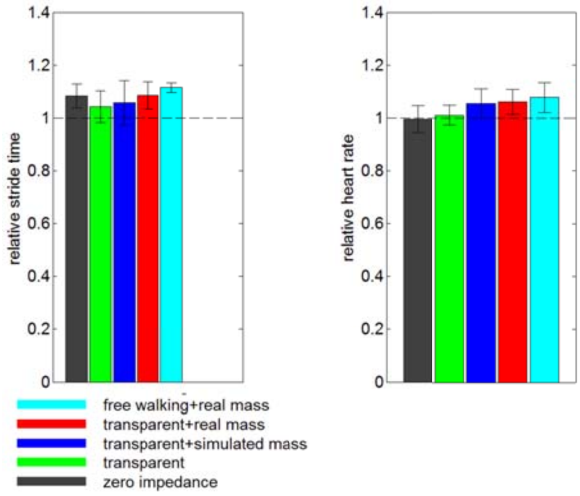

Figure 8. Mean relative stride times (left panel) and heart rate (right panel) for the different conditions. Stride times and heart rates are normalized to the free-walking trial and averaged across the subjects.

IV. DISCUSSION

The purpose of this pilot study was to show the feasibility of rendering the mechanical properties of different potential WR designs with LOPES. The first step in rendering the dynamics of any arbitrary WR is to compensate for the dynamics of LOPES itself. Although only three subjects have been measured, results from this study indicate that walking in LOPES results in an increase in EMG levels. This effect was acknowledged to a lesser extent in previous studies [8]. By using the inverse model of LOPES, EMG levels were reduced, suggesting that walking in LOPES is getting closer to free walking. However, they did not reach the level of free walking. The notion that the subject is less hindered by LOPES, when it is operated in the transparent mode, was also confirmed by a slight reduction in the interaction forces.

There are several possible explanations why EMG levels did not decrease to the level of free walking. First, it could be that the inverse model does not fully describe the dynamics of the LOPES. Currently, the model parameters are kept constant, but some parameters might change over the range of motion of the exoskeleton legs. Parameters like friction can change when the exoskeleton is repositioned. Furthermore, the estimated accelerations might be too low, resulting in an underestimation of the corrective torques. This can be the result of the selected cutoff frequency of the filter used to smooth the potentiometer signals, which might have been set too low. Finally, the corrective torques that must be applied to the knee joint are within the accuracy of the torque controller for the knee joint. Therefore, the knee joint is not contributing actively to the transparent mode.

The results of adding mass to the ankle during free walking are consistent with other studies performed during normal walking, although we found an increase of the stride time of 11 percent, whereas Browning [9] found an increase of 6 percent, with a 4 kg weight attached to the foot. This increase in stride time has been thought to be evidence that an energetically optimal stride time is selected on the basis of the pendular dynamics of the swing leg [10]. The increase in EMG patterns during walking with the mass are also in agreement with the literature in the sense that the muscles that initiate, propagate, and terminate leg swing generally increase their activation pattern. This is primarily caused by the fact that the weight increases the inertia around the knee joint.

Simulation of the mass resulted in a smaller increase in EMG, stride time, and heart rate, compared to walking in the transparent mode with the actual weight attached to the ankle. As mentioned above this could also be due to the filtering of the potentiometers. When the estimated accelerations are too small, the torques calculated by the inverse model of the WR will be too small, resulting in less torque that the subject needs to overcome to walk normally, and consequently a smaller increase in EMG. Another reason that might explain why simulating the mass tends to result in a smaller increase in EMG is that the LOPES can only apply the joint torques corresponding to the simulated mass. The vertical and horizontal forces at the hip, which are present when the subject carries the real mass, cannot be applied by LOPES.

V. Conclusion

In conclusion, the preliminary results presented in this paper suggest that LOPES can be used to render different WRs. However, more subjects must be included to see if the results are consistent and more effort should be put in retrieving proper estimations for the velocity and acceleration. The next steps will be to improve the model, study how subjects react to more complex WRs, and investigate if subjects can exploit the dynamics of intelligent and energy efficient WR designs, like the lightweight WRs with springs that store and release energy throughout the gait cycle.

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