Project full title:	**Evolving Morphologies for Human Robot Symbiotic Interaction**
Project Acronym:	**EVRYON**
Grant Agreement number:	**ICT-2007.8.5-231451**
Deliverable no.:	**D3.4**
Title:	**Evaluate the effect of different structural properties of the WR**

Contractual Date of Delivery	Project Month 22
Actual Date of Delivery	Project Month 24
Organisation Short Name of Deliverable Leading Partner	University of Twente (UT)
Organisation Short Name of Other Participants	TUD EPFL
Authors	Koopman B.
Editors	Asseldonk E.H.F., Kooij van der H.
External Review	
Version	1
WPs contributing to this deliverable	WP2 and WP4
Dissemination Level (*)	PP
Total number of pages (including cover page)	24

(*) Dissemination Level	
PU	Public
PP	Restricted to other programme participants (including the Commission Services)
RE	Restricted to a group specified by the consortium (including the Commission Services)
CO	Confidential, only for members of the consortium (including the Commission Services)

Project Abstract

The goal of the EVRYON project is to develop a novel approach for the design of Wearable Robots (WRs), e.g. exoskeletons, prostheses and other wearable mechatronic devices that can be used for a variety of applications, such as rehabilitation, personal assistance, human augmentation and more. Ideal solutions for such systems should aim at the optimal trade-off between performance, i.e. the level of assistance to be provided to the end-user, and some critical requirements, such as minimal weight and dimensions, low energy consumption and several other factors that can significantly affect the effectiveness and efficiency of WRs.

The basic idea behind the EVRYON project is that better WRs can be developed if the potentialities of 'embodied intelligence', and particularly of 'structural intelligence', are properly harvested and exploited.

EVRYON will develop an open-ended design process where both robot morphology and control are co-evolved and optimized in a simulation environment, where also the dynamical properties of the human body are taken into account. This approach is related to previous findings in the study of the emergence of structural intelligence in animals and artificial systems without any feedback control, such as preflexes in insects and emerging dynamic behaviours in passive walkers.

The EVRYON design methodology will originate a set of advanced tools for assisted mechatronic design, that will be validated by developing a novel prototype of a WR, i.e. an active orthosis for the lower limbs.

The EVRYON WR will integrate the kinematic, dynamic and control optimal solutions produced by the co-evolution process with additional variable impedance modules, which will allow the system to properly respond and adapt to the impedance patterns of human walking.

The WR prototype will be tested on a group of elderly subjects with age-related locomotion disabilities so to assess its acceptability and its ability to restore proper walking and increase personal autonomy.

Table of Contents

List of acronyms

The following abbreviations are used in this report:

LOPES	Lower Extremity Powered Exoskeleton
WR	Wearable Robot
AFO	Adaptive Frequency Oscillator
DOFs	Degree Of Freedom
SEA	Series Elastic Actuation
MIMO	Multi Input Multi Output
COM	Centre Of Mass
COP	Center Of Pressure
θ	Joint angle
$\dot{\theta}$	Joint angular velocity
$\ddot{\theta}$	Joint angular acceleration
τ	Joint torque
EMG	Electromyography
HR	Heart Rate
Hz	Hertz
SEA	Series Elastic Actuator

Executive Summary

This document illustrates how the effects of different structural properties of the Wearable Robot (WR) will be measured and evaluated using our gait trainer LOPES.

We recently introduced a new method that selectively cancels the dynamics of LOPES itself and adds the dynamics of the WR that needs to be evaluated. In this approach we make use of two inverse models. One consists of an inverse model of LOPES that is used to cancel its own dynamics, while the other inverse model represents the WR that needs to be simulated. 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. With this method different WR designs, that follow from the optimizations performed at EPFL, can be simulated on the LOPES. In a later stage (D3.5) the effect of different WR controllers can be added to this method.

Task 3.4 was supposed to end at project month 22, but a small delay must be recorded. The main reason for the delay is that the proposed method needs to be validated, and the time allocated for validation was not sufficient. Validation requires comparing walking in LOPES, while simulating a WR, to actually walking in that WR. To validate the LOPES as a tool to simulate the dynamical interaction with a WR, considering that no actual WR will be available before month 37, a preliminary simple example of a wearable robot has been considered, consisting of a mass attached to the ankle. In this way, EMG, heart rate, kinematics, etc. can be compared between the different conditions. Based on these measures we should be able to conclude whether simulating a WR has the same effect on the subject’s gait as actually walking in the WR. Although preliminary results seem promising, the proposed method needs to be fine tuned and validated on more subjects. After fine-tuning, this method will be used to model the wearable robot design that will follow from the optimizations performed at EPFL.

Another reason for the delay is that the selection of WRs, coming from the optimizations performed at EPFL, was finalized with a 2 months delay with respect to the workplan. EPFL’s optimization routines resulted in a selection of WR designs. To narrow down the number of WRs that need to be simulated on the LOPES, different selection criteria have been applied, thus obtaining a small number (2 or 3 out of an initial population of about 200 candidates resulting from optimization) of WR design candidates that can be simulated on the LOPES. While such selection was undergoing, fine-tuning of the simulations with LOPES has been finalized.

Preliminary results of this study, where an additional mass at the ankle was simulated, show that we are partially able to cancel the dynamics of LOPES. Additionally, the simulation of the mass at the ankle showed an increase in the muscles activity but not in the same order as during the control, where the subjects were actually carrying the mass. In conclusion, the results presented in this report suggest that LOPES can be used to render different WR. However more subjects need to be included to see if the results are consistent. Additionally, more effort should be put in retrieving proper estimations for the velocity and acceleration. This should improve the transparent mode as well as the modelled WR.

Note

We are now in the process of finalizing the simulation environment that is required to simulate the selected WR designs on the LOPES. This document consists of the results from the pilot study that shows the feasibility of rendering the mechanical properties of different potential WR designs with LOPES and an update on the current status of the simulation environment.

1 Introduction

1.1 Wearable robot design

In order 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. Although several effective wearable robot (WRs) have been developed for the lower extremities, more effective WRs and control strategies are still desired in order to realize natural and effective gait support.

During the last few years it has been successfully demonstrated that intensive, task-specific training can have a positive impact on the rehabilitation process for patients with central motor disorders [1, 2]. Additionally it is believed that treatment outcome could be further optimized by increasing the active participation [3, 4]. The latter can be achieved by adjusting the robotic assistance to the actual abilities and performance of the patient, so that the patient is only ’assisted as needed’ (AAN) [5, 6]. In order to provide AAN therapy force controlled gait trained are required [7], whereas the first generation of robotic gait trainers were mainly position controlled [8, 9].

Position controlled gait trainers enforce a specific gait pattern upon a patient by rigidly moving the legs through a prescribed gait pattern, whereas with force controlled gait trainers the amount of supportive forces can be adjusted to the capabilities and progress of the patient. These force (or impedance) controlled gait trainers, however, introduce new problems to be solved. Using this kind of control strategy means that the resulting motions will be the result of the provided assistance by the robot and the generated activity of the subject. When using low impedance values this can lead to unwanted motions or increase the likelihood that the robot and subject will start to walk out of phase [10]. In this perspective the development of patient friendly and intuitive controllers need special attention.

Another consequence of using impedance controlled gait trainers or WRs is related to its structural design. For force-controlled devices it is important to minimize the inertia of the device since control algorithms can only partly compensate for the inertia. Additionally the inertia of the device should be minimized in order to mimic walking outside the device as much as possible. Adding inertia to the patient’s legs can lead to adjusted kinematics by the patient, limiting the changes of transfer of the relearned gait tasks to overground walking.

For the new generation of rehabilitation gait trainers, or other supportive devices, it is therefore important that the mechanical design of the device allows free walking as much as possible. However, it is very difficult to predict how people will react and interact with the WR, as they might adjust their gait pattern to the mechanical restrictions or properties of the devise. In this report we propose a new method to simulate different WRs designs using the existing gait trainer LOPES.

1.2 Goal

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. This way different WR designs can be evaluated.

2 Methods

2.1 Subjects

Three healthy young subjects (2 males and 1 female, age: 25.7 ± 1.5 years, height: 1.81 ± 0.04 m, weight: 74.3 ± 11.5 kg), with no symptoms of neurological or orthopedic dysfunction, participated in this experiment. All subjects gave written informed consent to participate in this study. Table 1 gives the gender, age, height and weight of each subject.

Table 1 Subject information

Subject code	Gender	Age	Height (m)	Weight (kg)
A1	male	27	1.84	77
A2	male	26	1.82	92
A3	female	24	1.76	69

2.2 Experimental apparatus and recording

To render different mechanical WR designs LOPES (Lower Extremity Powered ExoSkeleton) was used. LOPES (see figure 1) is a treadmill-based lower-limb exoskeleton type robotic gait trainer. LOPES is impedance controlled and has 9 degrees of freedom (DoF), of which 8 are actuated (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. Figure 2 shows six of the nine DOFs that are integrated in the exoskeleton. It is torque controlled by means of series elastic actuation (SEA) [11]. Every DoF of LOPES is fitted with potentiometers that record the kinematics. Matlab xpc (Mathworks, Natick, USA) is used to control the applied torques by the exoskeleton joints at 1000Hz.

The used 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.

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

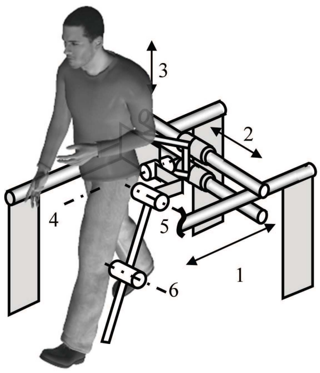

Figure 2 *Integrated Degrees of freedom of the LOPES exoskeleton. 1 2 and 3 represent peivic translations. 4,5 and 6 are hip adbuction, hip flexion and knee flexion respectively.*

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 6 DoF force sensors (ATI-Mini45-SI-580-20, ATI Industrial Automation, Goodworth, USA), which are positioned between the connection cuffs and the exoskeleton frame (see figure 3). The cuffs (Hocoma, Volketswil, Switzerland) used in LOPES are made of a rigid carbon fiber shell with Velcro straps and fixate the subject's legs to the robot. One cuff connects to the upper leg and two cuffs fixate 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, Autin, USA) and sent to the computer, where the data was stored for further processing.

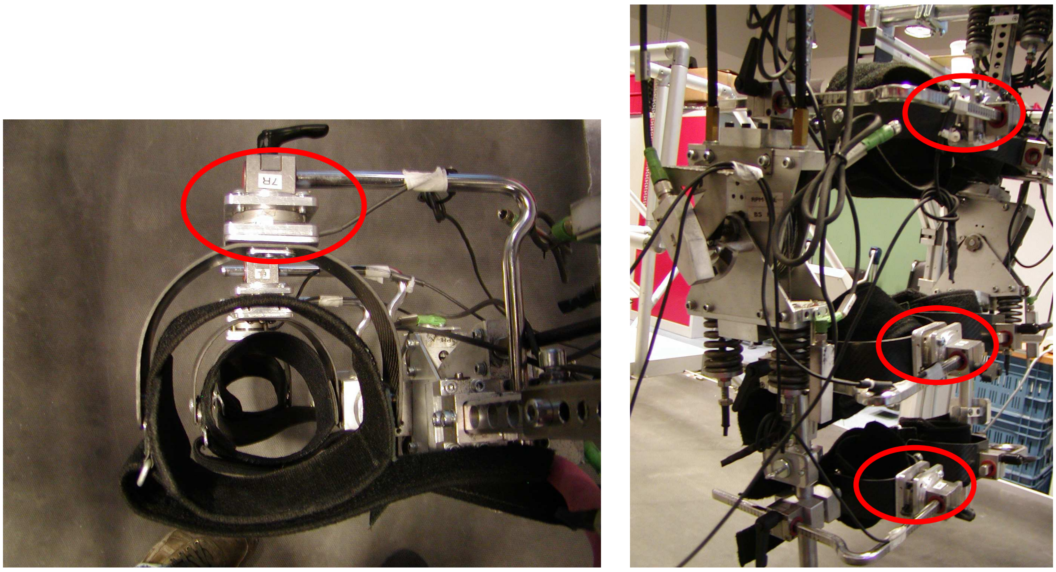

Figure 3 *6 DoF force sensors in combination with carbon shells and velcro straps tat fixate the upper and lower leg of the subject to the exoskeleton. 3 force sensors are positioned between the connection cuffs and the exoskeleton frame (red circles).*

As a predictor of the metabolic energy consumption heart rate was used. A wireless heart rate monitor (Polar, Oulu, Finland) was used to record the heart rate of the subject during all conditions.

During all trials muscle activation patterns were recorded by bi-polar surface electromyography (EMG) from the Rectus Femoris, Vastus Lateralis, Semitendinosis, Beceps Femoris, Tibialis Anterior, Gastrocnemius and the Gluteus Maximus. All recordings were performed on the left leg only. 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 was stored for further processing. A sinc signal was used to synchronize the EMG, LOPES and force sensor data.

2.3 Experimental design

Figure 4 shows the control strategy that is 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 its own dynamics. This is done by using a parameterized inverse model of the LOPES. In this model the mechanical behaviour 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, which represents the friction that exists in each joint. The parameters corresponding to the different LOPES segments are estimated using multi-input-multi-output (MIMO) system identification [12].

The input of the inverse model consists of the hip and knee angle, angular velocity and acceleration. At this moment LOPES is not fitted with accelerometers that measure the required inputs 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 are developed by Righetti, Buchli, and Ijspeert and can be used for various applications [13, 14]. The artificial oscillator that is used is based on an augmented Hopf oscillator. From a sinusoidal input this dynamical system is able to extract 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 especially the knee angle does not follow a sinusoidal pattern the adaptive oscillator is coupled to a non-linear filter, learning the trajectory envelop [15]. During preliminary experiments it is shown that this framework can easily be used to get an estimate of the joints position in the future. This way the delay caused by the filter can be compensated for by the oscillator, effectively creating a zero phase lag filter without the need for large signals buffers and creating smooth transitions when the phase lag changes.

This enabled the derivation of the angular velocity and acceleration purely based on the potentiometer signals. The states that are determined by the AFO serve as an input for the inverse model of LOPES and result in a torque command that is sent to the torque controller of LOPES. Using this setup 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. This inverse model will be used to calculate the torques that would have been exerted to the human when it would carry the WR. The resulting torques are again sent to the torque controller of LOPES.

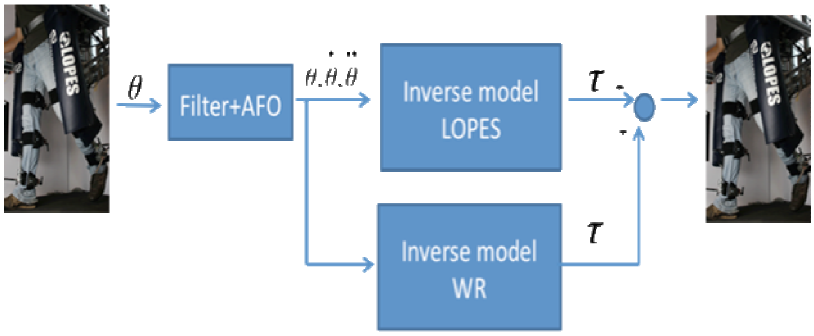

***Figure 4** Schematic representation of the proposed method to render the mechanical characteristics of a specific exoskeleton design on the LOPES. The inverse model of LOPES is used to eliminate the LOPES dynamics and the parallel inverse model of the WR is used to render the dynamic behavior of the WR.*

2.4 Experimental protocol

Before positioning the subject in the LOPES different antropomentric measurement were taken in order to adjust the exoskeleton segments lengths. Additionally the position of the cuffs was adjusted in 2 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, upper and lower shank were strapped to the exoskeleton. To let the subject become familiar with the device every subject walked for 5 minutes in the zero impedance mode with a constant speed of 3 km/h before testing began.

After this familiarization period different conditions were tested. Each condition was tested for 5 minutes with a 2 minute resting period in between. In table 2 the six different conditions are listed. 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 with an actual WR attached to the leg. Since this study is more about the proof of principle, and because of practical reasons, we choose 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 2 actual weights attached to his ankles. Two scuba diving belt lead weights, with a weight of 3.7 kg each, were used as point masses and fixed to the ankles using Velcro straps. In the “zero impedance mode” the subject walked in LOPES while no additional torques are exerted, whereas in the “transparent mode” the inverse model of LOPES is used to make LOPES more transparent, and consequently walking more natural. In the “transparent mode + simulated WR” condition the WR is simulated with the inverse model while the inverse model of LOPES is running in parallel. In the “transparent mode + real WR” the lead weights are used again as a representation of the WR.

All conditions are randomized to minimize the effects of fatigue. During all trial the subjects walked at 3 kph. The total protocol had a duration of 2 hours.

Table 2 List of tested conditions

Condition number	Definition
1	free walking
2	free walking + real WR
3	zero impedance walking
4	transparent mode
5	transparent mode + simulated WR
6	transparent mode + real WR

2.5 Data analysis

All signal processing was done with custom written software in Matlab (Nattick, USA). Gait kinematics recorded with LOPES were low-pass filtered with a 2nd order zero-lag Butterworth filter with a cut-off 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 2nd order zero-lag Butterworth filter (10 Hz). The raw EMG recordings were band pass filtered (10-400 Hz) with a 2nd order zero-lag Butterworth filter to remove movement artifacts, full wave rectified, and low-pass filtered with a low-pass 2nd order zero-lag Butterworth filter (5 Hz) to smooth the signal. All recorded signals, between the 4th and the 5th minute, were broken up into the individual stride cycles, based on the toe off 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 quantative measure by integrating the average activity of the stride cycle. The recorded interaction force was transformed into a quantative 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. The heart rate data was averaged between the 4th and the 5th minute for every condition.

3 Results

Figure 5 shows a typical pattern for the hip and knee angle recorded by the potentiometers (in red and green) and the constructed 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.

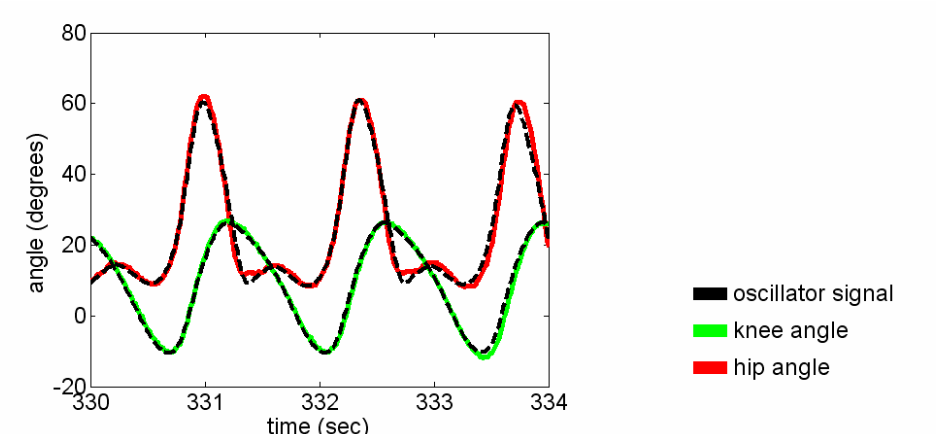

Figure 5 *Typical angle patterns for the hip and knee recorded by the potentiometers (in red and green) and the constructed AFO signal (black line) (subject 2, condition 3).*

In order to see if walking in the transparent mode becomes efficient the EMG levels of the zero impedance condition and the transparent condition are normalized to the EMG levels during the free walking condition (see figure 7). Figure 6 shows some typical EMG patterns for the different muscles during the free walking trial (dark pink), transparant mode (green) and zero impendace mode (pink), together with standard deviations. Note that some of the subjects show hardly any activity in the rectus femoris and vastus lateralis during the free walking trial. This explains the large relative activity and large standard deviation for these muscles in figure 7. Although so far only three subjects have been measured, we do see some trends. Figure 7 shows that walking in lopes increases the EMG levels above the level of free walking. In the transparent mode the EMG levels decrease, but do not reach the level of free walking yet. This suggests that walking becomes more efficient but not as efficient as free walking. This is also confirmed by a reduction in interaction force, as shown in figure 8. This figure shows the interaction force for the different sensorized cuffs during the transparent trial (normalized to the zero impedance mode). On average there is a reduction in interaction force for all cuffs, which indicates that the subject is less hindered by the exoskeleton.

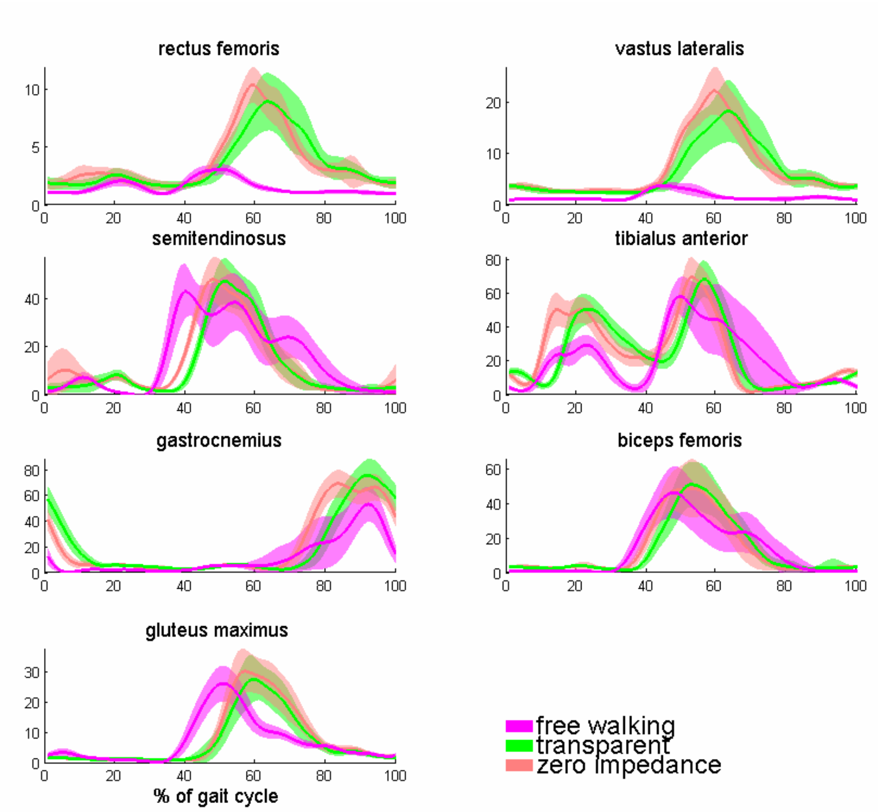

Figure 6 *Typical EMG patterns for the different muscles, as a percentahe of the gait cycle, during the free walking trial (dark pink), transparant mode (green) and zero impendace mode (pink) (subject 1, conditions 1,3,4) .*

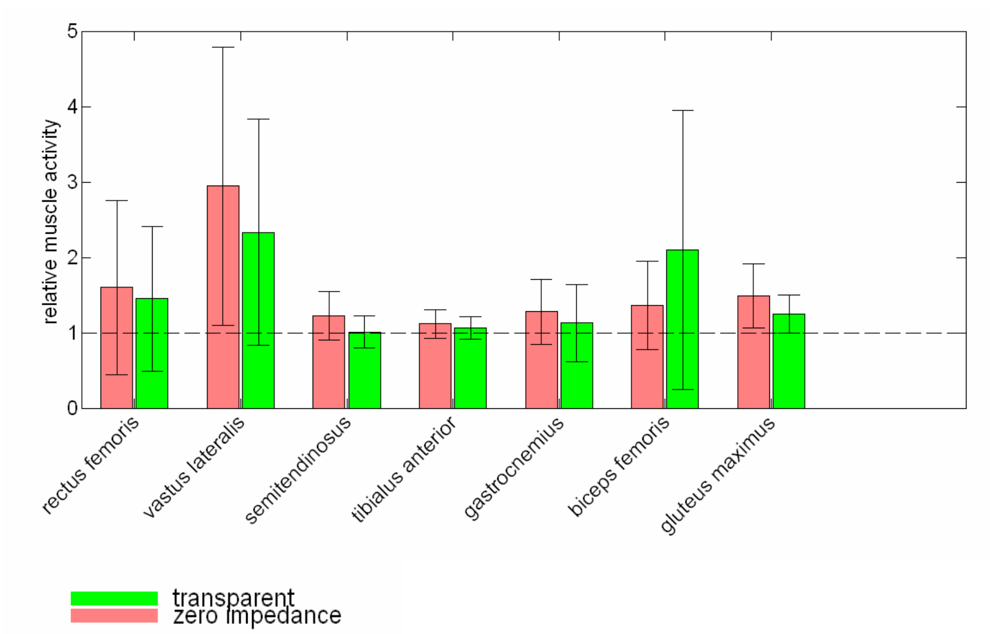

Figure 7 *Mean relative EMG levels for the different muscles for the zero impedance (pink) and transparent (green) mode. EMG levels are normalized to the free walking trial and averaged over the subjects.*

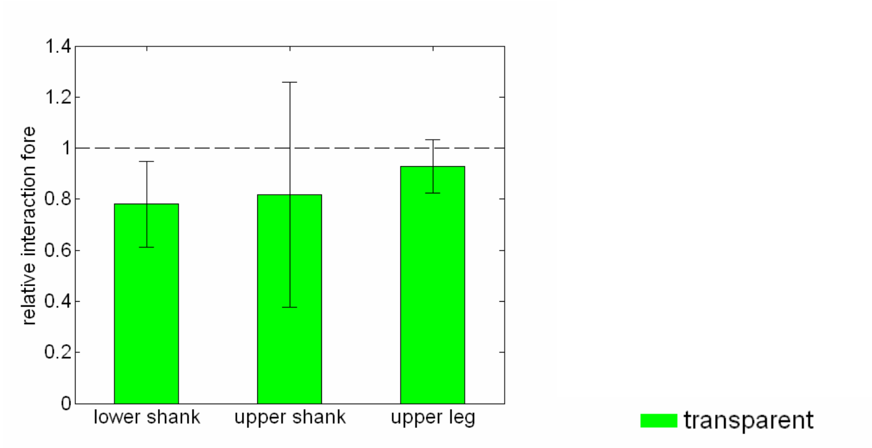

Figure 8 *Mean relative interaction force for the different sensorized cuffs for the transparent mode. Interaction forces are normalized to walking in the zero impedance mode and averaged over the subjects .*

To investigate the net effect of the added mass 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". Figure 9 shows that the EMG levels increase the most when the mass is applied during free walking. It also shows that modeling the mass at the ankle on top of the transparent mode increases the EMG levels (blue bars), but not to the level of actually carrying the mass (red bars).

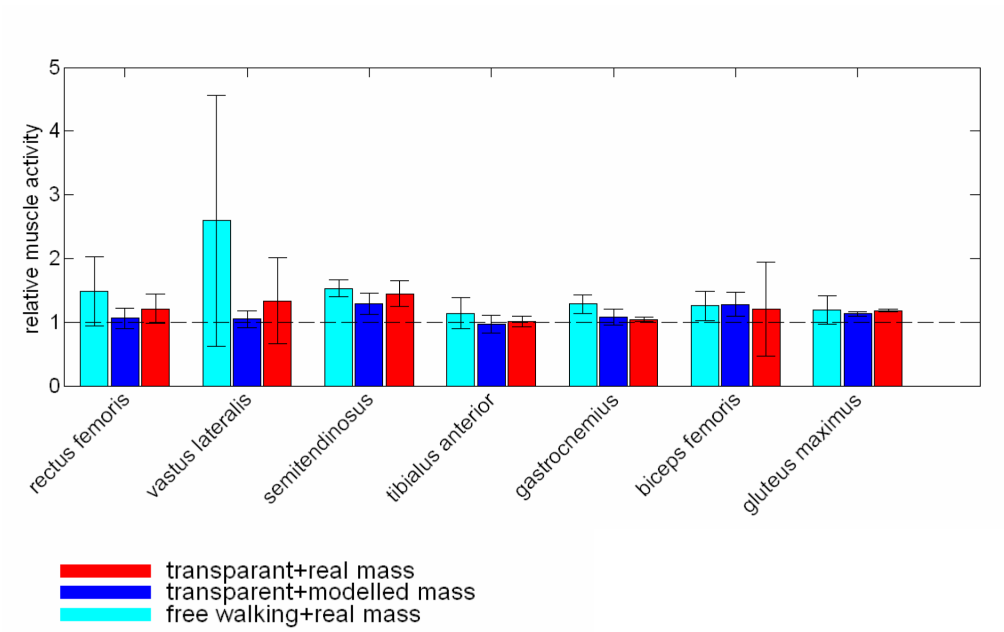

Figure 8 *Mean relative EMG levels for the different muscles for the transparent + real WR (red), transparent + modelled WR (blue) and free walking + real WR (cyan) trials. EMG levels are normalized to the transparent or free walking trial and averaged over the subjects.*

Figure 9 shows the stride times and heart rate for the different conditions (normalized to free walking). In the stride time figure similar trends as found in the EMG levels and interaction force can be observed. The transparent mode decreases the stride time more towards the free walking baseline and the added mass at the ankle has a larger effect when applied during free walking than during the "transparent + modeled WR" or "transparent + real WR" conditions.

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 modeled mass. Walking in the zero impedance and transparent mode does not seem to influence the heart rate.

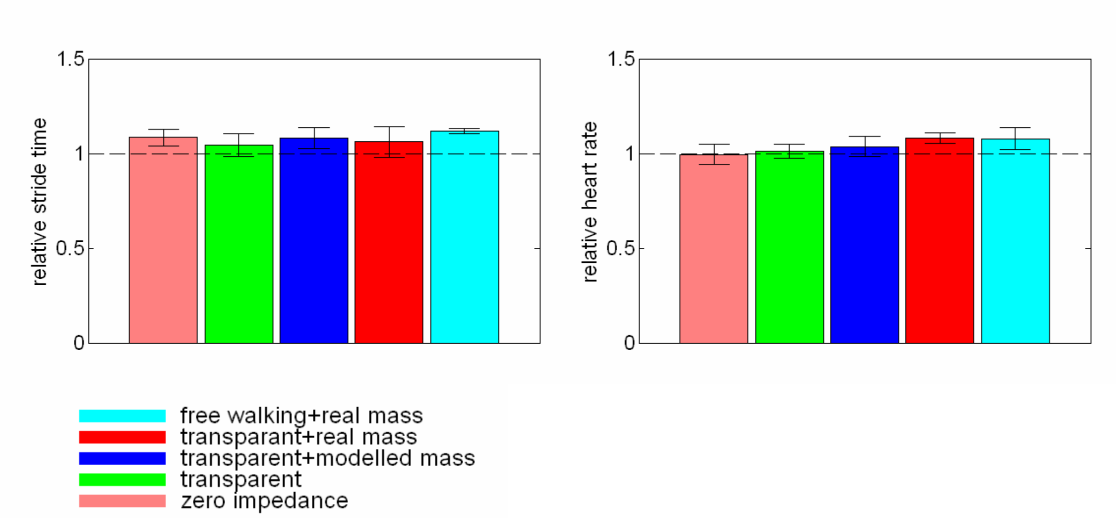

Figure 9 *Mean relative stride time (lef panel) and realtive heart rate (right panel) for the different conditions. Heart rate and stride time are normalized to the free walking trial and averaged over the subjects.*

4 Conclusion 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 so far the results from this study indicate that walking in LOPES results in an increase in EMG levels. This effect was acknowledged to a lesser extend in the previous studies [16]. With the inverse model of LOPES the EMG levels were reduced, suggesting that walking becomes more efficient, however they did not reach the level of free walking. The notion that the subject is less hindered by LOPES in the transparent mode was also confirmed by a reduction in the interaction forces.

There are several possible explanations why the EMG levels did not reduce to the level of free walking. First, it could be that the model parameters defined in the inverse model are not constant over the range of motion of the exoskeleton legs. Parameters like friction can change when the exoskeleton is positioned in another configuration. Furthermore, the cut off frequency of the filter used to smooth the potentiometer signals could be too low. This will result in a signal that is to smooth, resulting in an underestimation of the acceleration and consequently and underestimation of the corrective torques. Finally the torques that need to 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 a increase of the stride time of 12% whereas Browning [17] found an increase of 6% (with a 4 kg weight attached to the foot). This increase in stride time has been appointed to be evidence that an energetically optimal stride time is selected on the basis of the pendular dynamics of the swing leg [18]. 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 causes by the fact that the weight increases the inertia around the knee joint.

Modeling the mass tended to result in a smaller increase in EMG and heart rate compared to walking in the transparent mode with the actual weight attached to the ankle. This could also be due to the filtering of the potentiometers. When the estimated accelerations are to small the torques derived by the inverse model of the WR will be to small, resulting in less torque that the subjects needs to overcome to walk normally, and consequently a smaller increase in EMG

In conclusion, the preliminary results presented in this paper suggest that LOPES can be used to render different WR. However more subjects need to be included to see if the results are consistent. Additionally, more effort should be put in retrieving proper estimations for the velocity and acceleration. This should improve the transparent mode as well as the modelled WR.

5 Current status

We are now in the process of finalizing the simulation environment that is required to simulate the selected WR designs on the LOPES. Figure 10 shows the schematic representation of the proposed method to render a WR on the LOPES. This scheme is similar to the one described earlier (figure 4). Here the “inverse model WR” represents the dynamic structure of the WR that is being simulated and the “inverse model LOPES” block contains the inverse model of LOPES that is used to eliminate the LOPES dynamics. In this setup the inverse model of the WR consists of the dynamic structure of one of the optimized WR designs coming from the EPFL optimization. The dynamic structure is a combination of links and joints but also contain springs and dampers on the different joints. In contrast to the WR that was simulated during the pilot study, which only consisted of a mass at the ankle, this WR also included actuation. The Series Elastic Actuators (SEAs) of the WR are controlled to produce viscoelastic torques. This means that the amount of torque produced by the actuators depends proportionally on the deviation from a certain reference pattern (θ_{ref}) in terms of position and velocity. This reference pattern is an optimized reference pattern and will be synchronized with the user’s movement. These optimized reference patterns also follow from the EPFL optimizations. For each WR there is a specific reference pattern. . The stiffness (K) and damping (B) of the VIA also change during a gait cycle. The pattern in which they change is also optimized and will be synchronized with the user’s gait.

With this setup the different WR designs that meet the selection criteria can be simulated. During the simulations metabolic energy consumption, EMG patterns and gait patterns will be collected. In order to evaluate if the selected WRs are cognitively adequate and comfortable a questionnaire will be used. This questionnaire will contain general questions about the use of a supportive device like:

””Would you mind needing assistance for putting on or removing a supportive device”

”Would you be afraid of falling when using a supportive device”

but also more specific querstions about the WR that is being simulated like:

”Do you feel assisted by the wearable robot”

”Is the support synchronized with your movement when you change your cadance

Figure 11 shows a photo of a subject walking in the lopes while a certain WR is simulated. The visual display shows how the WR moves when the subject is walking.

To check if the experimental results are in agreement with the simulations performed at TUD and EPFL we will also test the WR designs on a dummy. This dummy will represent a passive human which has the same mechanical characteristics as in the ealier performed simulations.

As mentioned before, it is necessary to narrow down to 2 or 3 the total number of WRs to be simulated with the LOPES . To this aim, different selection criteria have been chosen during project month 24 so that simulations with the LOPES can start from month 25.

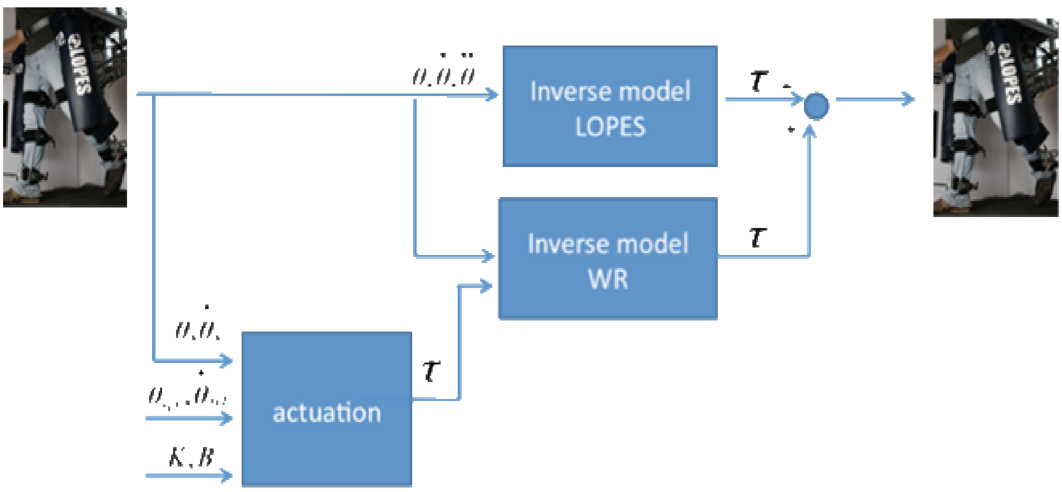

Figure 10 *Schematic representation of the proposed method to render the mechanical characteristics of a specific WR design on the LOPES. The inverse model of LOPES is used to eliminate the LOPES dynamics and the parallel inverse model of the WR is used to render the dynamic behavior of the WR. The actuation block represents the torques produced by the motors of the WR.*

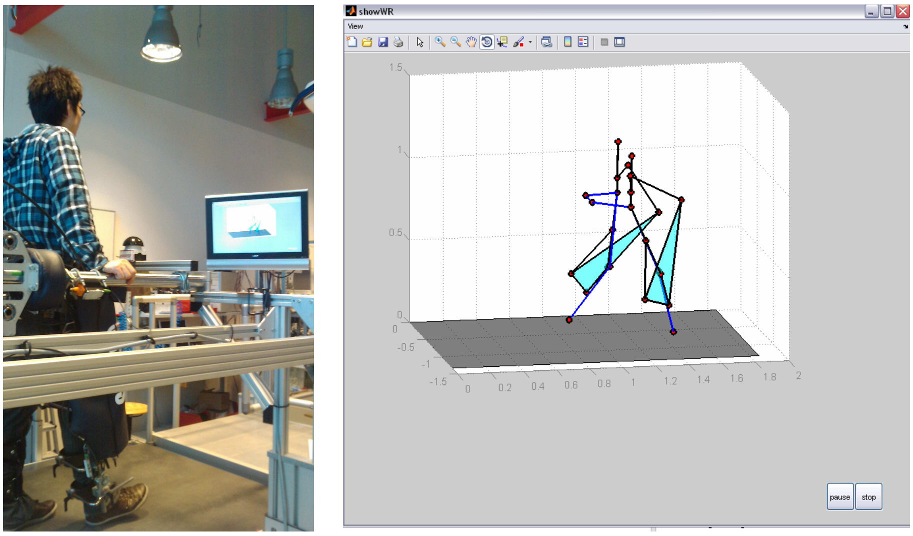

Figure 11 *Subject walking in the lopes while a certain WR is simulated. The visual display shows how the WR moves when the subject is walking. Its shows the LOPES frame in blue and the simulated WR in black and cyan.*

6 References

1. Kwakkel, G., *Impact of intensity of practice after stroke: issues for consideration.* Disabil Rehabil, 2006. **28**(13-14): p. 823-30.
2. Kwakkel, G., et al., *Intensity of leg and arm training after primary middle-cerebral-artery stroke: a randomised trial.* Lancet, 1999. **354**(9174): p. 191-6.
3. Lotze, M., et al., *Motor learning elicited by voluntary drive.* Brain, 2003. **126**(Pt 4): p. 866-72.
4. Perez, M., et al., *Motor skill training induces changes in the excitability of the leg cortical area in healthy humans.* Exp Brain Res, 2004. **159**(2): p. 197-205.
5. Emken, J., R. Benitez, and D. Reinkensmeyer, *Human-robot cooperative movement training: Learning a novel sensory motor transformation during walking with robotic assistance-as-needed.* J Neuroengineering Rehabil, 2007. **4**: p. 8.
6. Emken, J., et al., *Feasibility of manual teach-and-replay and continuous impedance shaping for robotic locomotor training following spinal cord injury.* IEEE Trans Biomed Eng, 2008. **55**(1): p. 322-34.
7. Veneman, J., et al., *Design and Evaluation of the LOPES Exoskeleton Robot for Interactive Gait Rehabilitation.* IEEE Trans Neural Syst Rehabil Eng, 2007. **15**(3): p. 379-386.
8. Colombo, G., et al., *Treadmill training of paraplegic patients using a robotic orthosis.* J Rehabil Res Dev, 2000. **37**(6): p. 693-700.
9. Hesse, S., D. Uhlenbrock, and T. Sarkodie-Gyan, *Gait pattern of severely disabled hemiparetic subjects on a new controlled gait trainer as compared to assisted treadmill walking with partial body weight support.* Clin Rehabil, 1999. **13**(5): p. 401-10.
10. Aoyagi, D., et al., *A robot and control algorithm that can synchronously assist in naturalistic motion during body-weight-supported gait training following neurologic injury.* IEEE Trans Neural Syst Rehabil Eng, 2007. **15**(3): p. 387-400.
11. Veneman, J.F., et al., *Design and Evaluation of the LOPES Exoskeleton Robot for Interactive Gait Rehabilitation.* IEEE Trans Neural Syst Rehabil Eng, 2007. **15**(3): p. 379-386.

12. Koopman, B., A. E.H.F., and H. van der kooij, *In vivo measurement of human knee and hip dynamics using MIMO system identification* 32nd Annual International IEEE EMBS Conference, August 31 - September 4, 2010, 2010.
13. Buchli, J., L. Righetti, and A.J. Ijspeert, *Frequency analysis with coupled nonlinear oscillators.* Physica D, 2008. **237**: p. 1705-1718.
14. Righetti, l., J. Buchli, and A.J. Ijspeert, *Dynamic hebbian learning in adaptive frequency oscillators.* Physica D, 2006. **216**: p. 269-281.
15. Gams, A., et al., *On-line learning and modulation of periodic movements with nonlinear dynamical systems.* Auton Robot, 2009. **27**: p. 3-23.
16. van Asseldonk, E.H., et al., *The Effects on Kinematics and Muscle Activity of Walking in a Robotic Gait Trainer During Zero-Force Control.* IEEE Trans Neural Syst Rehabil Eng, 2008. **16**(4): p. 360-370.
17. Browning, R.C., et al., *The effects of adding mass to the legs on the energetics and biomechanics of walking.* Med Sci Sports Exerc, 2007. **39**(3): p. 515-25.
18. Holt, K.G., J. Hamill, and A.R. O., *The force-driven harmonic oscillator as a model for human locomotion.* Hum. Mov. Sci, 1990. **9**: p. 55-68.