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BALANCE

Balance Augmentation in Locomotion, through Anticipative, Natural and Cooperative control of Exoskeletons

Deliverable 7.6

Robust and flexible interaction control adapted

for neurological impairment

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Summary

Lower limb exoskeleton assistive balance co-controller was adapted and tested with a stroke survivor whose lower limb (right leg) motor function was reduced. The controller was initially described and tested with healthy subjects in D4.4. In the tests with LOPES exoskeleton, the subject was walking straight while her pelvis was perturbed laterally on random occasion which required left/right hip abduction movements and lateral stepping for maintaining balance. The tests showed that exoskeleton hip abduction assistance helped the patient to recover balance and improved walking safety.

Contents

1 Human-exoskeleton co-control during balancing

This deliverable is a part of workpackage 7 of the BALANCE project. It is based on the controller design results of workpackage 4 (see D4.4) and the control parameters adaptation to the impairments observed in patients. We present the results of balance co-control testing for a patient. The tests were done with LOPES exoskeleton at Twente University, Netherlands. We investigated assisted balance recovery in walking with randomized lateral perturbation to the patient's upper body.

1.1 Background

Lower limb robotic exoskeletons have been proposed for human performance enhancement and neuromotor rehabilitation [10, 14, 19, 5]. However, improving the performance and safety of wearable robotic systems, and the development of novel functionalities, still remain challenging research topic. Gait assistance during walking is a main application for the majority of lower limb exoskeleton robots and multiple different control methods were proposed, such as: active impedance control to increase gait speed [9]; control of predefined gait patterns to support the weight shift during stepping [11, 6]; adjustable force fields to improve walking patterns in neurologically impaired users [19]; admittance control to shape the desired dynamic response in walking [16].

Overall only a few balance-assisting controllers in lower limb robotic exoskeletons have been investigated with human-participants: balance recovery with one [3] and with two subjects in [4]; influences on balance during walking in with passive exoskeletons in [2, 13]. Most importantly, the majority of the published studies did not consider human's active behavior in balancing co-control with the exoskeleton [15, 12, 17, 18]. More specifically, for stroke survivors no research has been done to see how their balance could be improved by support from a robotic exoskeleton. This deliverable reports an experimental study on human-robot balancing co-control during walking for a post stroke user. Currently it is unclear how the exoskeleton controller should be designed to act efficiently and cooperatively with its user in order to maintain balancing. We addressed this challenging question through the development and experimental evaluation of cooperative balance recovery control in a human-exoskeleton system. As a stable posture control during walking is essential, so wearable robotic systems should contribute to balance while taking into account the human user's motor response. We propose a simple balance recovery controller for a lower limb exoskeleton which can detect external body perturbations and provide assistance to the exoskeleton's user. More information on existing research is available in D4.4 (subsection 1.1).

1.2 Proposed balancing co-controller

The main idea of the proposed balance co-controller is to use natural human balance recovery and augment it with adapted assistance forces/torques produced by the lower limb exoskeleton. Hip abduction and stepping aside is a natural human response for balance recovery when relatively large lateral perturbations are applied to the upper body close to the centre of mass. The proposed controller augments this natural response by applying additional hip abduction torque with the exoskeleton electric motors.

Figure 1 shows the general balance co-control architecture which we describe and evaluate in this paper. We consider a lower limb robotic exoskeleton attached to its user at the pelvis and the legs. The general scenario considered is balance recovery from external perturbation applied at the upper body (pelvis level). In such situations, a stepping balance recovery strategy is employed depending on the magnitude of the pelvis perturbation.

Direction and stepping points (sometimes called capture points [21] or extrapolated center of mass [22]) will depend on the direction of the perturbation, resulting in stepping with left or right leg. This stepping balance recovery action will also depend on actual human-exoskeleton posture in standing or gait phase during locomotion (general case). We propose to implement an exoskeleton balance assistance controller that will take into account the natural human response and will assist the exoskeleton user with stepping for balance recovery once the perturbation is detected.

Perturbation detection can be implemented based on the exoskeleton robot state (its sensor measurements, for example centre of mass acceleration) but can also take into account the user's actual motion. The perturbation detection block can be also called as 'stability index' estimator as proposed in [8]. The controller should also take into account the current posture of the user in case of standing or gait phase during walking, so that the exoskeleton would act naturally. Finally, the assistance controller should also consider the actual mechanical interaction between the exoskeleton and the human's body to achieve efficient balance recovery cooperation and not to make interaction uncomfortable by constraining natural movements with the robot [20].

Robotic assistance. In the experimental study, an assistance torque was applied during certain trials after a pelvis perturbation was detected, to assist with balance recovery. The goal of the assistance was to apply external force to support the natural stepping behaviour yielding balance recovery (hip flexion). In unassisted pilot data, subjects showed an interaction torque associated with hip flexion. Because balance recovery is rapid, it is likely that the body uses simple and robust muscular control to regain balance. As such, simple torques were selected for assistance, with a filter applied to prevent rapid accelerations, which would be both unnatural and may disrupt natural motion control. Timing and magnitude of the assistance were determined by pilot testing, such that the torques occurred while the leg was in the air and participants had no complaints about interference from the assistance.

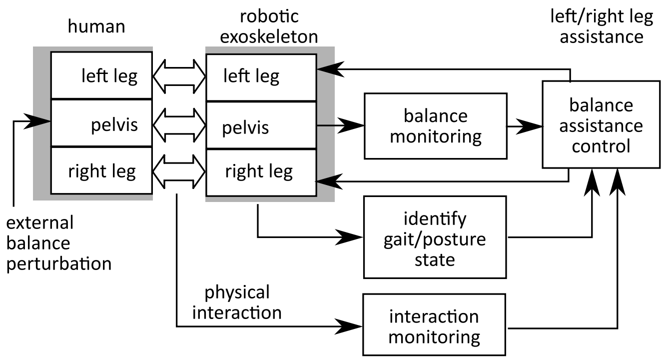


Figure 1: Exoskeleton-human balance co-control diagram. Force perturbations are applied to pelvis, while balance recovery assistance is applied to lower limbs by the exoskeleton.

Perturbation detection. Perturbations were detected by pelvis acceleration crossing a threshold of 2.5 m/s^2. Assistance was provided in the form of a 200 ms hip flexion torque to the leg performing a stepping action. The magnitude of assistance torque was set when the perturbation was detected, as

$$\tau_\circ = a_p + \Delta_p \dot{a}_p, \tag{1}$$

where a_p is the pelvis anterior-posterior acceleration and acceleration prediction time interval $\triangle_p$=50 ms. Therefore, the assistance torque magnitude was determined by the pelvis acceleration threshold, but the jerk factor predicted the acceleration over the next 50 ms which acted to estimate the perturbation magnitude. The assistance hip flexion torque was applied for 200 ms with a 5 ms low-pass filter, as shown in the left panel of Figure 2.

The applied assistance torque τ_{assist} also decayed exponentially when a large net work was produced:

$$\tau_{\text{assist}} = \begin{cases} \tau_\circ & \text{if perturbation detected} \\ \tau_\circ \mathrm{e}^{-\gamma t} & \text{after } W(t) > 0.01\,\text{J} \\ 0 & \text{if } t > 200\,\text{ms or no perturbation} \end{cases} \tag{2}$$

The net mechanical work $W(t)$ at time t was defined as the difference between the cumulative positive and negative mechanical work at each time step from the perturbation onset 0 to t:

$$W(t) = \int_0^t max\{P(\tau), 0\} d\tau - \int_0^t min\{P(\tau), 0\} d\tau, \tag{3}$$
$$\text{with P} = \omega_{\text{hip}} \cdot \tau_{\text{assist}}$$

where ω_{hip} is the hip abduction angular velocity. In this calculation of work, assistance torque τ_{assist} is used instead of interaction torque because of the noise inherent in the measured value, which could result in large variations in detected work. A 5 ms low-pass filter was applied to τ_{assist} in order to limit large changes in forces applied to the subject.

2 Balance co-control test with stroke survivor

2.1 Experimental setup

The LOPES III lower limb robotic exoskeleton used in the experiments [7] (Fig. 2) is composed of leg and pelvis attachments with horizontal push/pull rods connected to robotic shadow legs, an actuation mechanism with electric motors and controllers, and a treadmill. The shadow legs can be actuated in shank flexion/extension, thigh flexion/extension and abduction/adduction. The actuation mechanism enables pelvis control in forward/backward and mediolateral directions. Additionally, LOPES III is equipped with adjustable body weight support for a user. The exoskeleton's passive ankle joints are connected to the leg guidance bars with a series of revolute joints with axes intersecting at the ankle joint. The distances between hip, knee and ankle joints is adjustable for each user to enable natural and comfortable movements. The exoskeleton is operated in admittance control at 1000 Hz. During the experiments LOPES actuators were operated in haptically transparent mode with the pelvis anterior-posterior active degree-of-freedom (DoF) used for force perturbations and right thigh movements (hip joint) used for balance recovery assistance. The experimental design was implemented in Simulink (MathWorks, Natick, MA, USA) on top of the LOPES low-level controller.

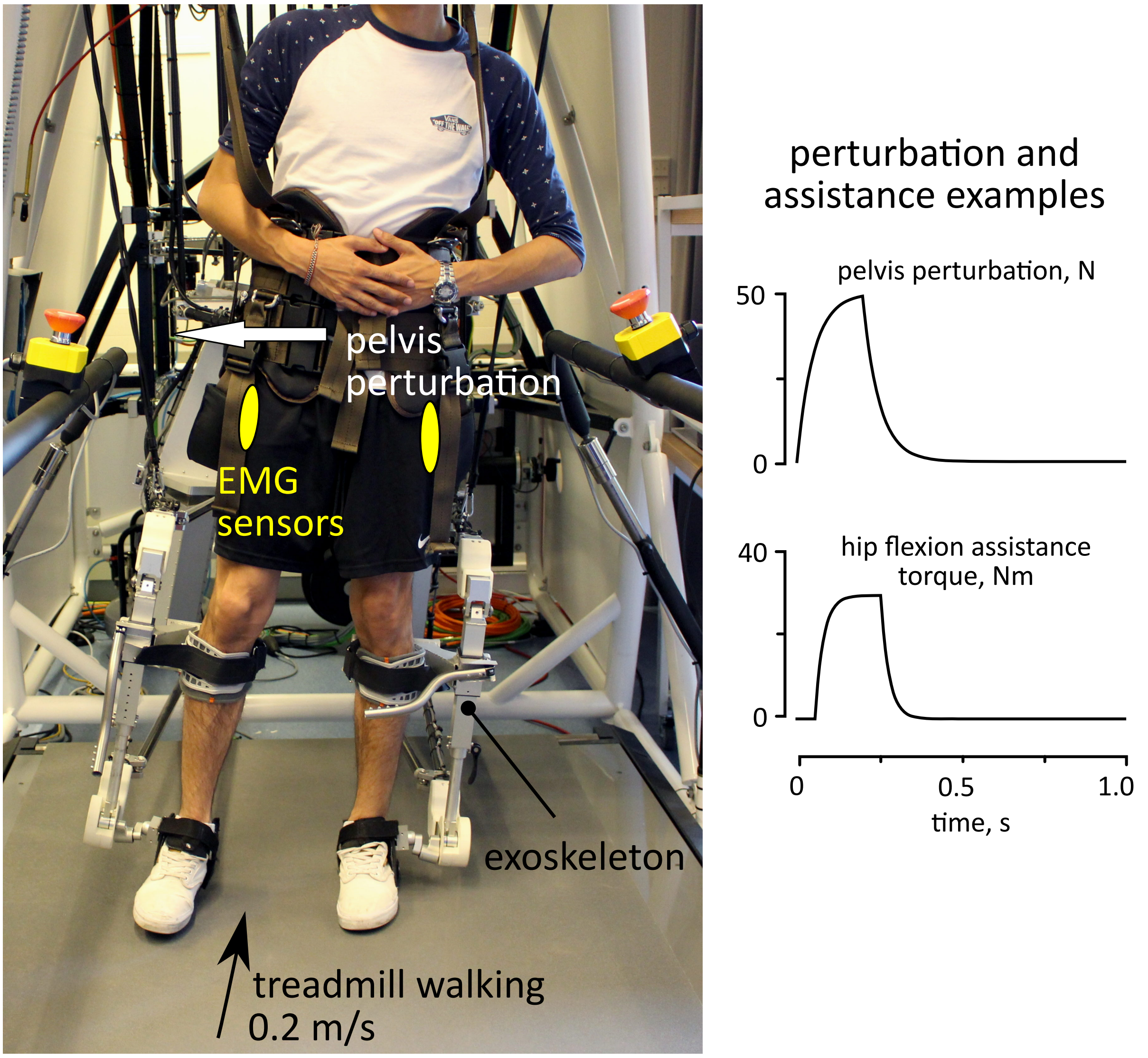


Figure 2: Experimental setup. A human-subject walking on the treadmill. Pelvis is randomly perturbed laterally and exoskeleton assists left/right hip abduction to recover balance. Left plots demonstrate sample pelvis perturbation and hip abduction assistance profiles are shown. Note, that a photo of a healthy human-subject is used for this illustration.

2.2 Patient-participant

Two patients were tested in the setup. However, due patient-specific difficulties in adaptation to the lower limb exoskeleton, we excluded one of the participants. The results described in this report are from a 71 years old post stroke female subject with reduced mobility of the right leg (body mass 61 kg, height 167 cm).

2.3 Protocol

This subject walked on a treadmill at 0.2 m/s. 200 ms lateral perturbations at the pelvis were applied at the beginning of the swing phase of walking, randomly every 2.5–4 gait cycles. Gait phase was computed by the LOPES controller. The perturbation levels were significantly reduced in patient trials relative to healthy subjects: it was set empirically to approximately 10-15% of body weight with a 50 ms low-pass filter and were tested before the experiment with each subject to cause loss of stability without excessive discomfort or triggering the LOPES safety triggers for high acceleration. The subject was instructed to continue walking normally when the perturbations occurred. She was instructed not to use arms and handrails during the test, but no physical restrictions were implemented. In addition, electromyography (EMG) was recoded from the right and left gluteus medius muscles, which are responsible for hip abduction (using the Delsys Trigno wireless EMG system).

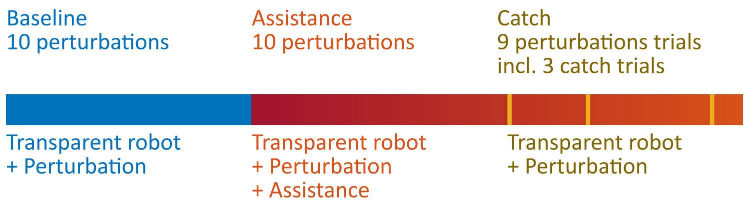


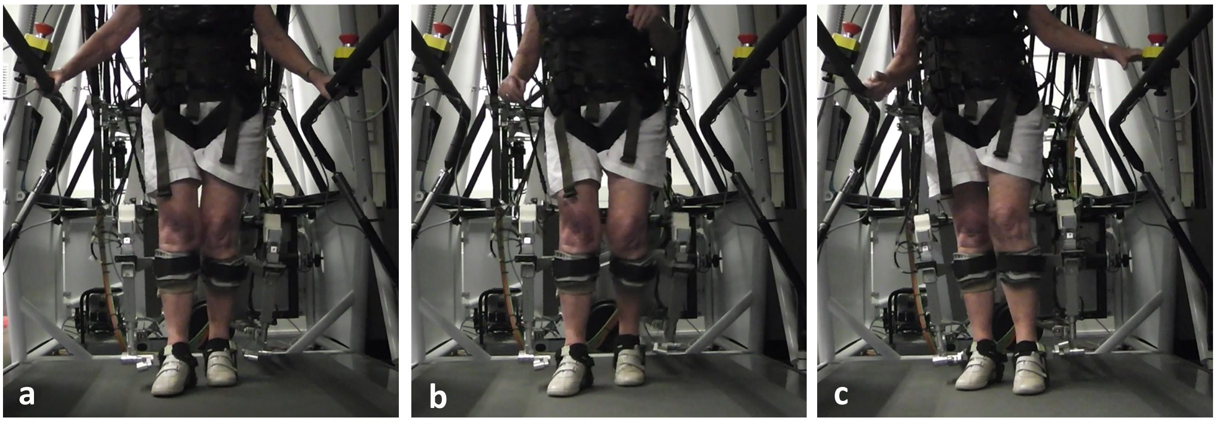
Figure 3: Experimental protocol for the test.

Procedure: Prior the main test, the subject was given the opportunity to just walk in LOPES robot without any assistance and we determined the preferred walking speed and made sure that they could walk in the device without using their hands. This was done several days before the balance recovery tests with pelvis perturbations. During the actual test, the subject first walked in the exoskeleton for 2–3 minutes with no perturbations. This was followed by 10 *Baseline* perturbations, where no assistance was provided. Next was the *Assistance* stage, where 10 perturbations were applied with robotic assistance. The test was concluded with 3 catch trials in which balance recovery without assistance interspersed with 9 assistance trials. The total length of this procedure was approximately 5 mins. The protocol is represented schematically in Figure 3. This test was repeated four times for the patient and patient had sufficient amount of resting time between the repetitions. The experimental protocol was approved by the ethical research committee.

3 Results

Pelvis perturbation magnitude was adjusted for each testing trial. At first, 45 N pelvis perturbation were tested, however it was later increased to 60 N to trigger stronger balance recovery response in patient (for information, 150-220 N perturbations were used in the experiments with healthy subjects). Importantly, in the beginning of the test the patient had to use handrails to support walking even when no pelvis perturbations were applied (See Figure 4a). During the first trial, the handrail was used to maintain stability after the perturbation in 60% of the cases overall and no difference for assisted and non-assisted cases was observed. Improvement in balance recovery was observed during the second trial: handrails were used in 70% of the cases when no assistance was used, and only in 10% of the cases when robotic balance recovery assistance was applied. In most of the cases with the robotic assistance it was sufficient to use stepping strategy to recovery balance (See Figure 4b). Decrease in the number of cases when handrail was necessary to maintain balance after the perturbation indicates that the participant was able to adapt to perturbations and robotic assistance. During the third trial handrail support was used only in 20% of the cases without robotic assistance, and was not used when robotic assistance was applied. Similarly, no handrail support was required in the fourth trial when robotic assistance was present. In all trials, handrails support was required at least in 2 out of 3 perturbation cases for catch trials (See Figure 4c).

Compared to the experiments with healthy subjects, in the tests with the patient we had to significantly reduce the level of perturbations to make balance recovery feasible. Usage of handrails were required to maintain balance in some trials even if robotic assistance was applied compare to the observations from healthy subjects. Importantly, no modifications of the robotic assistance controller for patient scenario were required. Same approach for hip abduction assistance was used both for healthy user and patient scenarios, demonstrating the flexibility and practicality of the proposed control strategy. Compare to healthy subjects a lower pelvis acceleration threshold was used to detect the perturbations. Further investigation on the patients gait kinematics and muscle activation is required for the detailed analysis (this will be reported in a future publication).



with handrail use | without handrail use | handrail use in catch trial

Figure 4: Patient during experiment (captured from video recording). **a:** walking with handrail support during initial trials; **b:** adaptation to walking without handrail support, including trials without and with exoskeleton assistance; **c:** usage of handrail support after pelvis perturbation in a catch trial.

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