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BALANCE

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

Deliverable 4.4

Learning standing and walking behaviours

in human/robot collaborative control

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Summary

In this deliverable, the human behaviour and co-control for balancing with an exoskeleton during standing and walking were investigated experimentally. The experiments were performed with the LOPES exoskeleton at University of Twente, Netherlands. In standing tests, four participants were submitted to a perturbation, consisting of a forward force impulse applied to their pelvis, that forced them to step forward with the right leg for balance recovery. Trials with and without exoskeleton assistance to move the stepping leg's thigh were conducted to investigate the influence of the exoskeleton's control assistance on balancing performance and a potential adaptation. Analysis of the body kinematics and muscle activation demonstrated that robotic assistance modified the stepping leg trajectories by increasing hip and knee flexion, increased reaction speed and decreased the step duration, increased biceps femoris and rectus femoris muscle activity (in 75% of participants). It was easy to use and did not require learning. In the second experiment seven participants were asked to walk on a treadmill while wearing the LOPES exoskeleton. Similar to the standing experiment, the participants' pelvis was randomly perturbed left/right forcing the subject to step in the corresponding direction to maintain stable walking. Muscle activation (gluteus medius) was significantly reduced for all subjects when robotic balance recovery assistance was used. All participants were able to adapt to robotic hip abduction assistance easily. Assisted movements were not disrupting movements and no motor learning was required to adjust the participant's motor control.

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Chapter 1

Standing and balancing behaviour in human-robot co-control

1.1 Introduction

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 topics. 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].

While some of the previous works [6, 9] state that modification of human gait characteristics improves stability in walking, very few results were published on improving balance function with the help of a wearable exoskeleton. Balance control in standing and walking is a crucial exoskeleton control function, which should be considered in the development and evaluation of wearable robots for posture and gait assistance [1].

Existing works on augmenting balance with robotic exoskeletons have important limitations. In [12, 17] a lower limb exoskeleton with variable stiffness actuators was used for balance recovery, however experiments were performed with the robot only. Human inspired balancing strategies were proposed in [15] for balance rehabilitation and in [18] for able body assistance, but the proposed approaches were not tested in a coupled human-robot system. In [2] the influence of passive exoskeleton mechanics on human biomechanics of walking and balancing was investigated. It was shown that a passive exoskeleton degraded the balancing performance when perturbations were applied to the exoskeleton's user.

The importance of the exoskeleton's allowing pelvis anterior movements in improving stability of walking was demonstrated in [13]. Studies [12, 13] indicated that the exoskeleton design, specifically its kinematic structure, can significantly degrade balancing function. In [3] a knee exoskeleton is used for balancing assistance based on mimicking estimated human knee joint impedance control. The experiments were conducted with one participant only and limited data was provided. A powered ankle-foot orthosis from [4] was used to improve standing balance for two spinal cord injury patients.

Overall only a few balance-assisting controllers in lower limb robotic exoskeletons have been investigated with human-participants: balance recovery with one [3] and two [4] subjects during stance; influence of passive exoskeleton on balance during walking [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].

The present work reports an experimental study on human-robot balancing co-control in standing. Currently it is unclear how the exoskeleton controller should be designed to act efficiently and cooperatively with its user in order to maintain balancing during standing. 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.

We conducted an experimental study in which a participant's pelvis was perturbed and balance recovery action with stepping was performed with and without robotic assistance. The goal of the study was to systematically investigate human-in-the-loop exoskeleton co-control. Movement and muscle activity analysis of four participants demonstrated how human neuromotor control can adapt to balance assistance torques generated by the exoskeleton. The results of the experiments may be useful to indicate how lower limb exoskeletons can be controlled to work efficiently and safely with their users.

1.2 Balance co-control in standing

Figure 1.1 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 [23] or extrapolated center of mass [24]) 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 the 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 natural human motor responses 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 also be called as stability index estimator as proposed in [25]. The controller should also take into account the current posture of the user in case of standing or gait phase during walking, so that

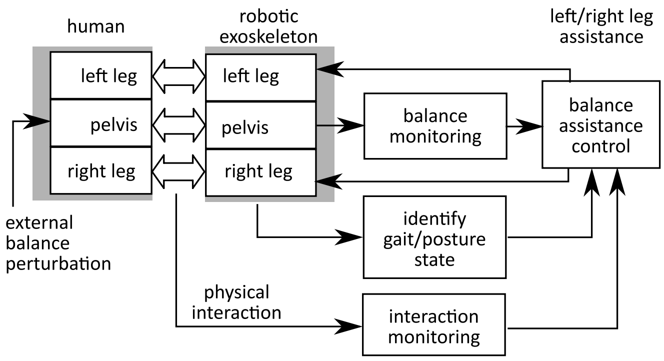

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Figure 1.1: Exoskeleton-human balance co-control diagram. Force perturbations are applied to pelvis, while balance recovery assistance is applied to lower limbs by the exoskeleton.

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 [26].

Robotic assistance. In the experimental study during certain trials, an assistance torque was applied after a pelvis perturbation was detected, to assist with balance recovery. The goal of the assistance was to apply external force supporting natural behaviour in stepping required for 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 unsafe 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.

When a perturbation was 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, with magnitude

$$\tau_\circ = a_p + \Delta_p \dot{a}_p, \tag{1.1}$$

where a_p is the pelvis anterior-posterior acceleration and the acceleration prediction time interval is Δ_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 low-pass filter (cut off 200 Hz), shown in the bottom of Figure 1.2b.

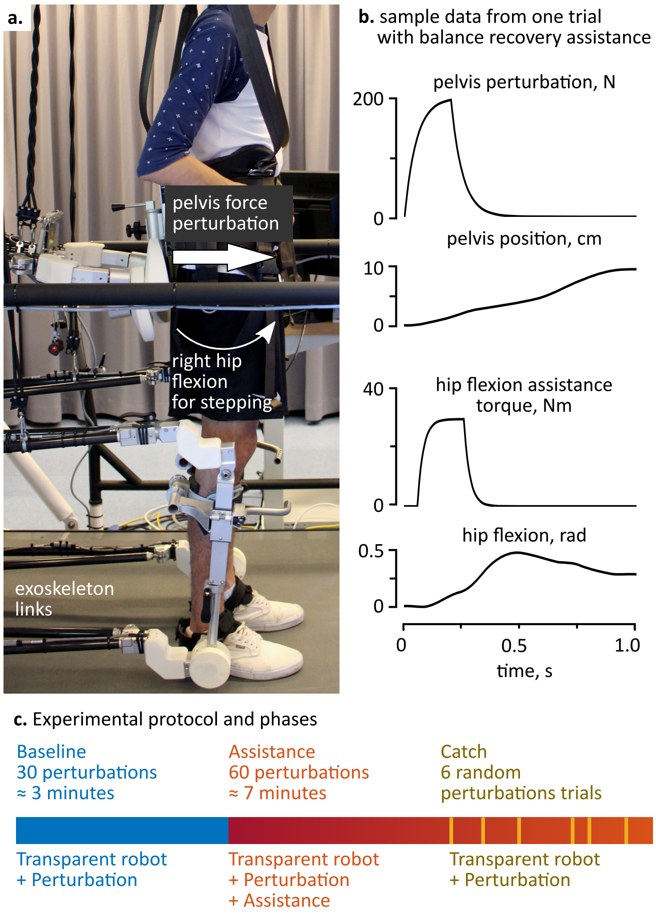

Figure 1.2: Experimental setup and protocol: **a:** Participant before stepping, interacting with LOPES III exoskeleton; **b:** Sample single trial recorded data for balance recovery assistance; **c:** Experimental protocol.

1.3 Experiment

1.3.1 LOPES III robotic exoskeleton

The LOPES III lower limb robotic exoskeleton used in the experiments [7] (Fig. 1.2a) 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 controlled in admittance at 1000 Hz [7]. During the experiments LOPES actuators were operated in

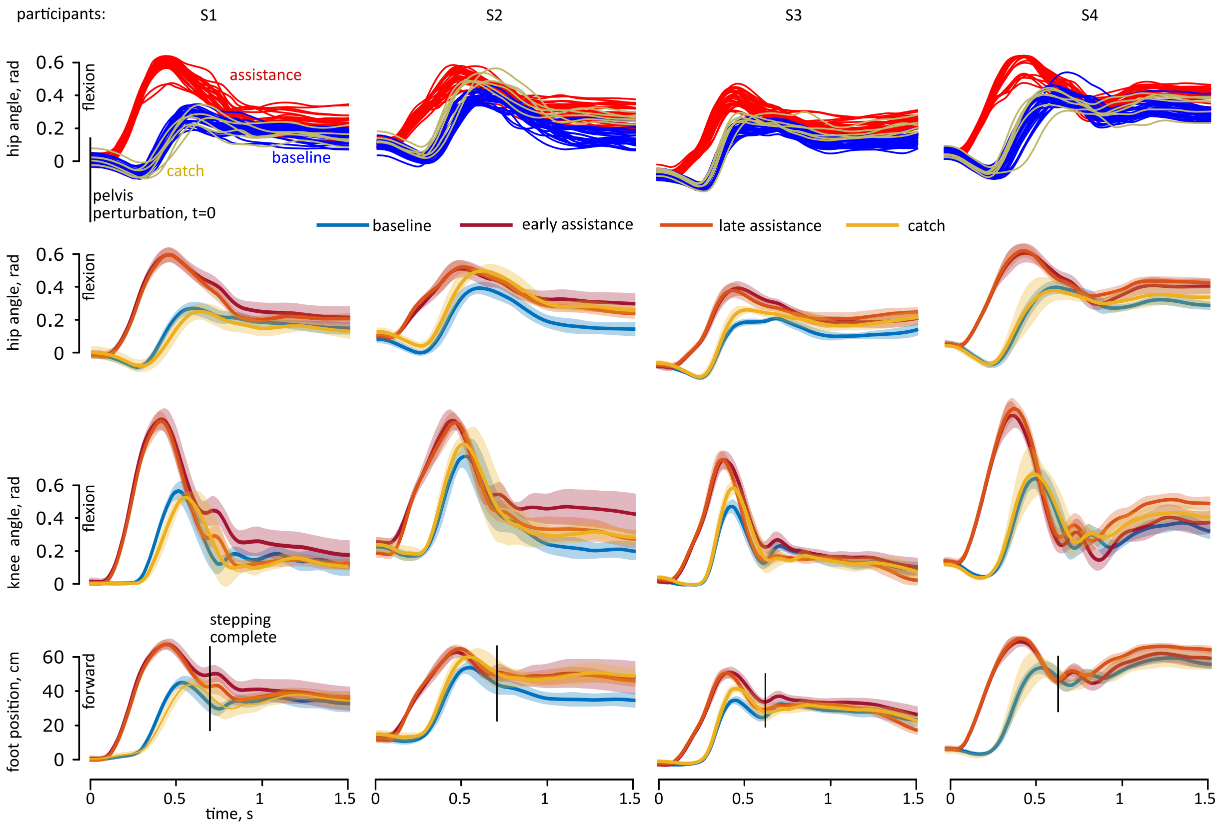

Figure 1.3: Right leg kinematic data during stepping for all participants (S1-S4). Black vertical line in the last row corresponds to completion of stepping.

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.

1.3.2 Protocol and participants

Four healthy adult participants (1 female, 3 males; age 25–35) took part in the study which was conducted at University of Twente, Netherlands, and was approved by the institutional ethical committee. In the beginning of the experiment the participants stood upright with their feet together and weight slightly shifted toward the left leg, as shown in Fig. 1.2a.

A 200 ms forward perturbation (step filtered with a low-pass filter with 50 ms time constant) (Fig. 1.2b) was applied randomly every 4 to 10 s (uniform distribution), requiring subjects to take a step with the right leg to regain stability. The magnitude of the perturbation was approximately 35% of body weight. This was determined by initialisation trials preceding the experiment procedure to cause loss of postural stability without causing discomfort. The participants were instructed to hold the harness and not use their arms for balance recovery. After each step, subjects returned to an upright stance and stood straight for the next perturbation.

The experiment was composed of three phases shown in Figure 1.2c: 1) baseline: balance recovery without robotic assistance, 2) assistance: balance recovery with robotic assistance (split into two subphases: early and late assistance), 3) catch trials: balance recovery without assistance interspersed with assistance trials. Before the experiment the participants familiarised with the system, which took around 2-3 minutes per participant. The goal of the baseline experimental stage was to record balance recovery behaviour after pelvis perturbations when the exoskeleton was haptically transparent and no assistance was used. In total 30 pelvis perturbations were introduced in the baseline phase. The baseline phase was followed by an assistance phase of 60 trials during which robotic assistance was applied to support stepping and balance. Six catch trials with robotic assistance removed were randomly introduced during last 30 trials of the assistance phase. The catch trials were used to investigate whether there was any learning in the assistance phase. The overall experiment took around 15-17 minutes per participant.

1.3.3 Data processing and statistics

The experimental data recorded at 1000 Hz included joint angles and participant-exoskeleton interaction torques for hip and knee flexion/extension; pelvis position in horizontal plane, and centre of pressure (CoP) measured from the treadmill. Interaction torques were estimated by the exoskeleton's low level real time controller.

The *stepping duration in balance recovery* was calculated using the CoP. A threshold was set at 20% of the maximum lateral centre of pressure velocity in the full duration of the experiment. The *end of each recovery step* was defined as the time point after the perturbation at which the centre of pressure lateral velocity crossed this threshold.

The *forward step length* was computed as the mean foot position in the 200 ms following the end of each step with respect to the pelvis position. The data analysis showed that this time was sufficient for stepping after the pelvis perturbation was applied. Pelvis position along the anterior-posterior axis was set to zero at the onset of each perturbation, and the right foot x position was calculated using the recorded right hip flexion (θ_{HF}) and knee flexion (θ_{KF}) angles and the estimated leg segment lengths:

$$x_{\mathrm{foot}} = x_{\mathrm{pelvis}} + l_{\mathrm{thigh}} \sin\theta_{\mathrm{HF}} + l_{\mathrm{shank}} \sin\theta_{\mathrm{KF}}. \tag{1.2}$$

The peak right hip interaction torque magnitude (maximum of the absolute value) was taken from the perturbation onset to the step duration.

Electromyography (EMG) recordings of muscles on the right leg associated with the hip flexion/extension movement (gluteus maximus, biceps femoris, rectus femoris, and gastrocnemius) were obtained at 1000 Hz and saved during balance recovery with Delsys Trigno wireless EMG system. The rectus femoris muscle is involved in hip flexion, while gluteus maximus and biceps femoris muscles contribute to hip extension. Additionally, the gastrocnemius muscle activation was recorded to observe any influence of the assistance controller to ankle-foot plantar flexion and knee flexion.

Raw EMG signals were de-meaned, high-pass filtered at 50 Hz, rectified, and low-pass filtered at 10 Hz (4th order Butterworth). Signals were normalised by subject, such that the highest peak value was 100% for each leg over all trials. This allowed for comparison between subjects.

Within-subjects and between-subjects statistics were similarly computed for the standing experiment, with a one-way ANOVA comparing *baseline*, *early assistance*, *late assistance*, and *catch trials*. Because of the smaller number of perturbations in each condition, early assistance consisted of the first 15 trials of the condition, while baseline and late assistance were the last 15 trials of their respective conditions.

Recorded data was truncated, grouped by experimental stage into trials and every trial corresponded to a single right leg stepping action. Each trial recording consisted of 3000 data points per recorded parameter.

Outliers were removed prior to data analysis, based on the right hip flexion/extension recordings within each experimental state. Specific trials with hip trajectories markedly different from the mean over all trials were considered as outliers and eliminated. This was detected from the root-mean squared error (RMSE) relative to the overall mean, with a threshold of two times the average RSME. Furthermore, trials in which the peaks of at least one of the EMG recordings exceeded the double standard deviation margins around the mean recorded values were also eliminated from the analysis. This was done to exclude the trials with abstruse EMG recordings which could be caused by body movement artifacts. It was verified that no more than 10% of trials per experimental stage were removed for each participant.

1.4 Results

We analysed the right leg movements, muscle activation, pelvis and centre of pressure kinematics in stepping during balance recovery for each participant per experimental phase. The results per participant and experimental phase are presented as follows: right leg kinematics in Figure 1.3, pelvis movement and centre of pressure in Figure 1.4, right leg muscle and activity in Figure 1.5. In these figures the recordings are aligned in time, such that the time instant t=0 corresponds to the moment when a forward force perturbation was applied. We have also computed the within-participant and the between- participant statistical significance in the data, as shown in Table I and Figure 1.6, respectively.

1.4.1 Hip flexion

The first row of Figure 1.3 shows the time history of right hip flexion in all trials for each participant (columns S1-S4) after removing the outliers. Once a pelvis perturbation was applied a participant had to flex his/her right hip in order to move the right foot forward and recover balance with stepping. Hence, in all participants we see an increase of hip flexion during approximately the first 0.5 s. Stepping was complete at approximately 0.7 s, when the right foot was placed on the ground and hip angle was static. For all participants a clear difference in trajectories between baseline (no assistance, blue lines) and assistance (red lines) is observed. The recordings in catch trials (shown in yellow) are close to the baseline trajectories. The second row of Figure 1.3 shows the mean (bold lines) and standard deviation margins (in shadow) for the right hip flexion recordings in four experimental stages (shown in four colours). Assistance torque applied to the hip during the assistance trials increased the hip flexion magnitude compared to the baseline cases. Hip extension movement which is observed during first 200-300 ms after the perturbation in the baseline trials is not present in early and late assistance trials. Hip flexion in catch trials is similar to the baseline trials.

Within participant analysis showed that for all participants the mean hip flexion with assistance was significantly larger than in baseline or catch trials (Table I). For all participants, no statistical difference was observed between baseline and catch trials. Between participant analysis showed larger hip flexion in early and late assistance than in baseline trials, which was statistically significant (p <0.05), as shown in Figure 1.6a. There was no significant difference between baseline and catch trials. There was also no significant difference in hip flexion between early and late assistance trials in all participants.

1.4.2 Knee flexion

Right knee flexion recordings for each participant are shown in the second row of Figure 1.3. Similar to hip movements, we observe knee flexion during 0.5 s after the perturbation, after which it extends to approximately 0.3-0.4 rad when the stepping is complete. Even though there was no direct assistance applied at the knee joint level, the thigh movements were similar to the hip movements. In the assistance trials, knee trajectories were larger, than in the baseline and the catch stages of the experiment. Knee flexion in the catch and the baseline trials do not differ from each other significantly.

Table 1.1: Within-participant statistical analysis (T-test, significance for p <0.05).

Variable	Experimental stage	# of significant participants		
		<	=	>
mean hip flexion	Baseline vs. Early	4	0	0
	Baseline vs. Late	4	0	0
	Catch vs. Early	4	0	0
	Catch vs. Late	4	0	0
	Early vs. Late	0	4	0
mean knee flexion	Baseline vs. Early	4	0	0
	Baseline vs. Late	4	0	0
	Catch vs. Early	4	0	0
	Catch vs. Late	4	0	0
	Early vs. Late	1	1	2
step duration	Baseline vs. Early	0	1	3
	Catch vs. Early	0	0	4
	Catch vs. Late	0	0	4
	Early vs. Late	0	4	0
step length	Baseline vs. Early	3	1	0
	Baseline vs. Late	2	2	0
	Baseline vs. Catch	0	4	0
integrated biceps femoris EMG	Baseline vs. Early	1	3	0
	Baseline vs. Late	3	1	0
	Baseline vs. Catch	0	4	0
	Catch vs. Early	0	3	1
	Catch vs. Late	2	2	0
	Early vs. Late	1	3	0
integrated rectus femoris EMG	Baseline vs. Early	3	0	1
	Baseline vs. Late	3	1	0
	Baseline vs. Catch	0	4	0
	Catch vs. Early	3	0	1
	Catch vs. Late	3	1	0
	Early vs. Late	3	1	0

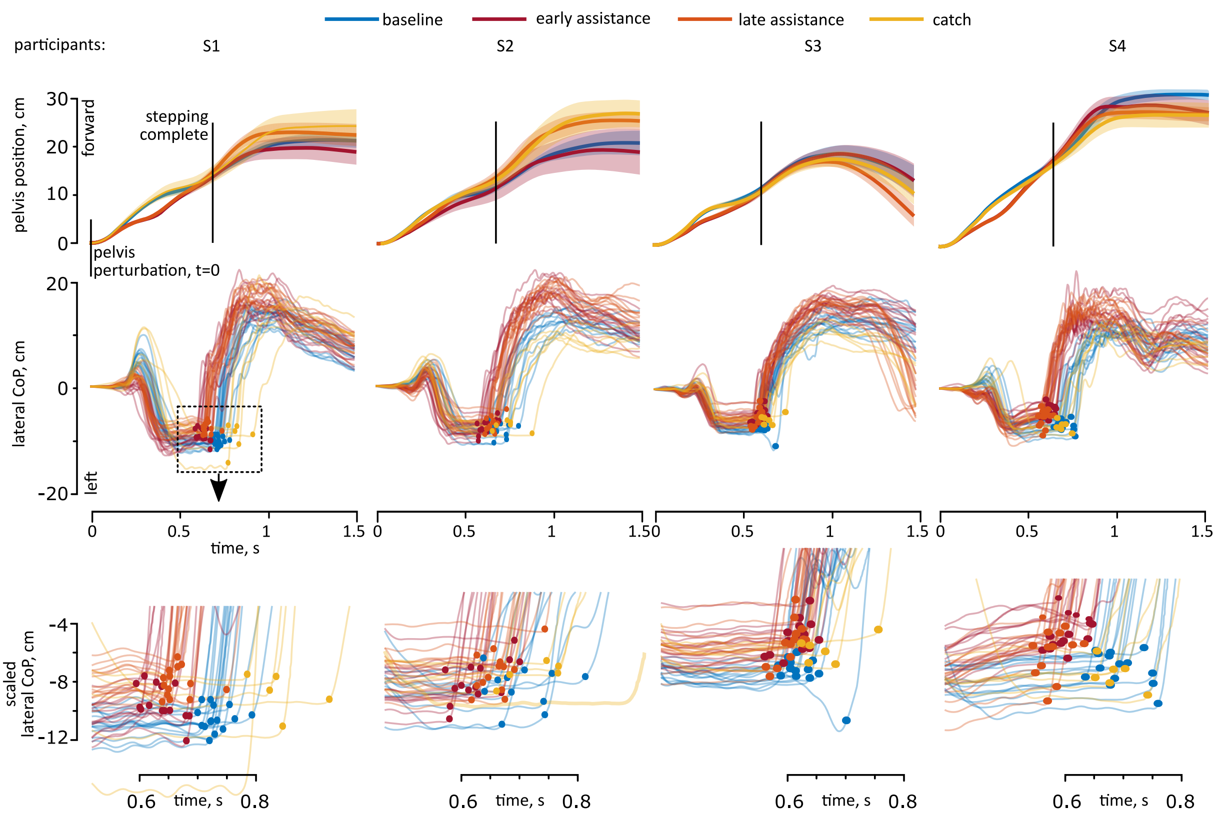

Figure 1.4: Pelvis movement forward and lateral center of pressure during stepping for all participants (S1-S4). Individual trials are shown with markers indicating the end of the step.

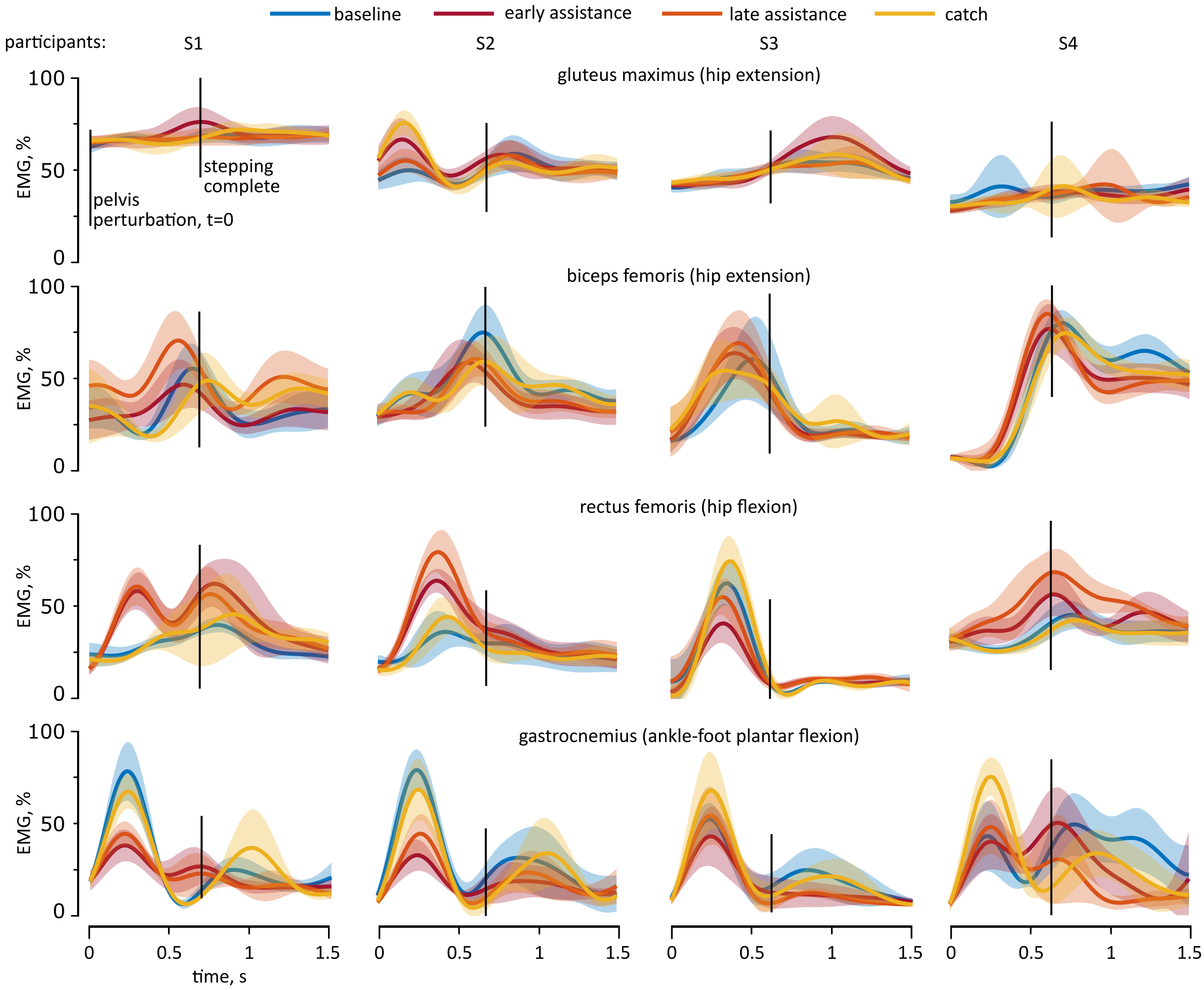

Figure 1.5: Normalised right leg muscle activity recordings during stepping for all participants (S1-S4).

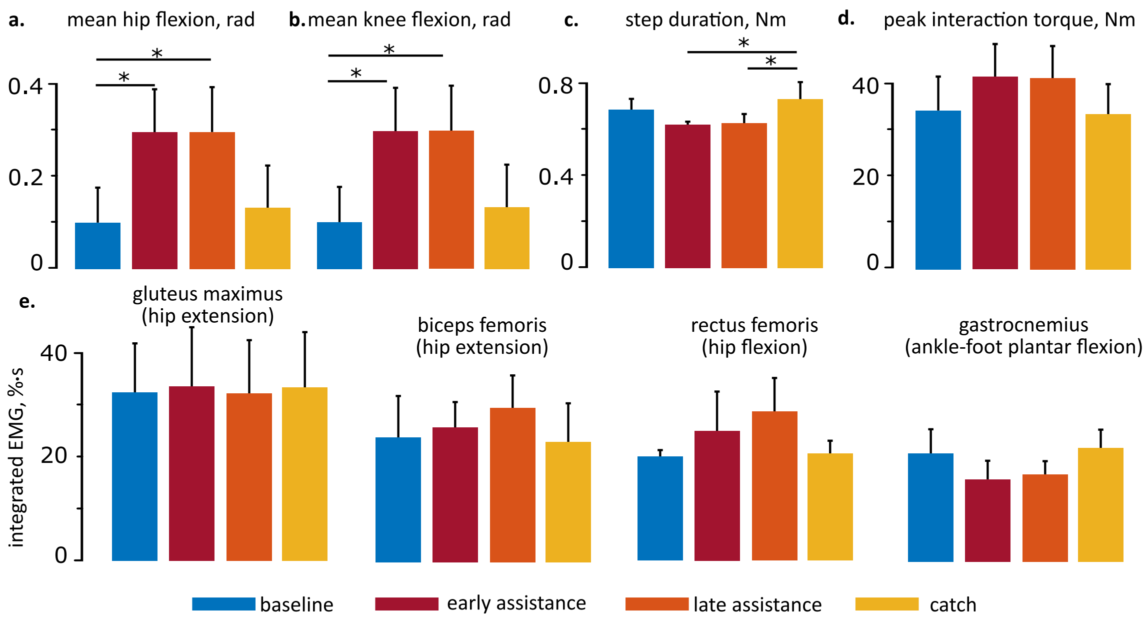

Figure 1.6: Results of the between-participant statistical analysis for right leg stepping: **a-c** - mean kinematic parameters for all participants; **d** - mean exoskeleton-participant hip interaction torque; **e** - mean of normalised integrated EMG for all participants.

Similar to hip flexion, within participant analysis showed that for all participants the mean knee flexion was significantly increased in assistance trials compared to baseline and catch trials (Table I). For three subjects there was no difference in knee flexion in baseline and catch trials. Results of between participant analysis of knee flexion are shown in Figure 1.6b. It showed that knee flexion was increased significantly in all assistance trials (p <0.05).

1.4.3 Foot placement, step length and duration

Calculated trajectories for right foot position in the forward direction are shown in the last row of Figure 1.3. During the assistance trials, the right foot was placed further away in conjunction with increased hip flexion and knee extension when compared to baseline and catch trials, which is clearly observed for the participants S1, S2 and S4. Within-participant analysis showed that the step length was increased in the early assistance trials compared to the baseline recordings for three participants. For all participants, there was no significant difference between the baseline and catch trials (See Table I). Between participant analysis revealed no significant differences in step length between experimental stages ($p = 0.15$, $F = 2.14$).

Within participant analysis showed that step duration was longer in catch trials than in early and late assistance for all participants (Table I). There was no significant difference in step duration between early and late assistance trials. Three participants had reduced step duration in early assistance compare to baseline trials ($p < 0.05$). Between-participant analysis showed that step duration was significantly increased in catch trials compare to assistance trials ($p < 0.05$), as presented in Figure 1.6.

1.4.4 Pelvis movement and CoP

Pelvis forward movement trajectories and CoP lateral displacement are shown in Figure 1.4. The pelvis moved forward once the force perturbation was applied until the stepping for balance recovery was complete, with the body slowing and stopping at 15-30 cm distance from the origin. As seen from the first row in Figure 1.4, the recordings of pelvis position during stepping are characterised with small standard deviation for participants S1, S3 and S4. For the same participants, the pelvis movement forward in the assistance trials was slowed down from around 200 ms when right hip assistance torque was applied. Probably, slower pelvis movements were the result of the reaction torques caused by fast hip flexion movements.

Lateral CoP trajectories with markers indicating the end of stepping are shown in the second row and the scaled up version is shown in the third row of Figure 1.4. Once the perturbation was applied and the right leg was lifted from the ground for balance recovery the CoP quickly deviated to the left, until the stepping was complete. Once stable double stance was achieved the CoP moved to the right as the overall centre of mass of the body was deviated to the right as well. For all participants except for S2, it is the clear to see that stepping was complete faster compare to baseline and catch trials (see third row of Figure 1.4). Also, the CoP lateral deviation to the left in the assistance trials was slightly smaller for the participants S1, S3 and S4.

1.4.5 Muscle activity and interaction torque

The normalised muscle activation of the right leg during stepping is presented in Figure 1.5. The between subject statistics for integrated muscle activation over the stepping duration per participant per experimental stage during stepping is presented in Figure 1.6e.

There was no effect of experimental stage on integrated gluteus maximus contraction found in the between-participants ANOVA ($p = 0.99$, $F = 0.02$), shown in the first row of Figure 1.5.

The biceps femoris EMG is shown in the second row of Figure 1.5. There was no difference in integrated biceps femoris EMG between conditions in the between-participants ANOVA ($p = 0.55$, $F = 0.74$). However, within-participant analysis showed that three subjects had increased biceps femoris muscle contraction from baseline to late assistance, and for all participants there was no significant difference between the muscle activation in baseline and catch trials (Table I).

The third row of Figure 1.5 shows activation of rectus femoris muscle. There were no significant between-participant effects of experimental stage on integrated rectus femoris muscle contraction ($p = 0.11$, $F = 2.49$). However, within-participant analysis showed that three participants had increased rectus femoris contraction in assistance trials compared to baseline and catch trials; for all participants there was no difference of activation in baseline and catch trials; three participants showed an increase from early to late assistance (Table I).

The last row of Figure 1.5 shows activation of gastrocnemius muscle. There was no main effect of experimental stage on gastrocnemius contraction in the between-subject ANOVA ($p = 0.10$, $F = 2.65$).

As shown in Figure 1.6e normalised muscle activation did not depend on the experimental stage based on the between-participants analysis. Similarly, there was no significant between-participants effect of experimental stage on the peak human-exoskeleton hip joint interaction torques as presented in Figure 1.6d (p=0.25, F=1.57).

1.5 Discussion

This report presented an experiment in using lower limb robotic exoskeleton to augment human balance in standing. To our knowledge, few studies have investigated human-exoskeleton balance co-control performance [2, 4, 3, 13]; these contain single-subject experiments with few measurements. Due to the complexity of the experimental setup used in this work and the variability of human motor behaviour, only some of the observations were statistically significant, and it is thus not straightforward to interpret the results. However, our study demonstrated that simple robot controllers can be efficiently used for assisting balance in complex physical interactions between a user and robotic exoskeleton. It also demonstrated important behavioural trends which we briefly discuss in this section.

Hip flexion assistance generally did not change step length, although within-subject results suggested limited adaptation to the assistance. We interpret this as a positive result, as the study was conducted with healthy subjects who can already take appropriately sized steps to recover balance, and the assistance provided did not negatively impact this ability. Steps duration was slightly longer without assistance (catch and baseline trials) than in assisted trials, indicating that subjects adapted their behaviour in anticipation of the assistance, and completing the recovery step took longer when the expected assistance was absent.

Unsurprisingly, the hip flexion assistance torque resulted in greater average hip flexion during the step. Notably, this appears to have occurred as a result of the hip flexion occurring earlier, rather than simply causing a larger peak flexion due to the presence of external assistance torques. This occurs because the assistance torque was applied 55 ms following the perturbation, as opposed to the reaction time delay of up to 100-200 ms for the subjects' muscle activation. Interestingly, a similar increase in knee flexion was observed in assistance trials even though the robotic assistance was applied only to hip movements. This indicates that it is possible to apply robotic assistance only to selected DoFs of the human body and the (rest of the) body will naturally adjust balance recovery response through neuromechanical interaction.

In wearable robots control design, applying torque on the body could result in uncomfortable or even dangerous interaction with the user. Our experiments showed that with the proposed balance recovery assistance there was not a large increase in human-exoskeleton interaction torques for hip and knee joints. There is evidence of a marginal mean increase in interaction torques in assistance trials, which was however not significant.

Muscular contraction in the rectus femoris and biceps femoris (hip flexion/extension movement) often increased as a result of assistance. However, there was no significant difference in the between-participants analysis likely due to high between-subject variability. The within-subjects differences underlined the importance of selecting appropriate assistance torque magnitudes and timing parameters. Interestingly, the between-participants analysis also showed a trend for reduced gastrocnemius muscle activity (ankle-foot plantar flexion movement) indicating that less ankle activity is required to push off for stepping. Even though this decrease was statistically not significant, it demonstrates that the participants' whole body neuromuscular control was actively adapting to exoskeleton assistance forces. The gluteus maximus muscle does not appear to have been significantly involved in the movement or impacted by the assistance. As can be seen from the single subject results in Figure 1.5, contraction of the muscle occurred after the step was complete, which was not considered in the analysis.

Overall, the effect of assistance on muscular activation is mixed. While the gluteus maximus appears to not be involved or affected by the assistance, activation of biceps femoris and rectus femoris (both muscles in the thigh) apparently increased, and gastrocnemius muscle contraction decreased. The gastrocnemius is the only muscle that showed a reduction of contraction and is also the only muscle measured that is not responsible for hip movement. The rectus femoris and biceps femoris are antagonistic muscles, but both showed a tendency toward increasing contraction; added assistance led to increased stiffening through co-contraction of the muscles responsible for hip flexion and extension.

Importantly, the within-participant analysis showed that the proposed stepping controller for balance recovery did not require any motor learning from the users. Statistical comparison of early and late assistance trials did not reveal any differences and balance recovery performance remained same. It was demonstrated that simple assistance control can be easily applied to collaborative human-robot balance recovery task. A user's neuromuscular motor control could rapidly adapt to the forces and movements produced by the lower limb exoskeleton robot and no motor skills learning was required. However, the limitations of our study include the limited number of participants and the balance recovery scenario which only considered forward perturbations in standing.

The results presented in this report are crucial for understanding human-robot interaction in critical situations such as balance recovery. The results of this study will be used to develop novel collaborative balancing controllers for lower limb robotic exoskeletons that interact seamlessly with natural behaviours.

Chapter 2

Walking and balancing behaviour in human-robot co-control

2.1 Introduction

Walking exoskeletons are often used for human performance enhancement and for neuromotor rehabilitation. However very few of such systems are equipped with balance control functionality [1, 18, 16]. So far only preliminary studies [3, 4] have examined how the exoskeleton controller should be designed to act efficiently and cooperatively with its user in order to maintain balance during standing and walking. We addressed this challenging question through the development and experimental evaluation of cooperative balance recovery control strategy in a human-exoskeleton system. Same controller as described in section 1.3 for standing was used for walking balance recovery assistance.

2.2 Balance co-control in walking

We adopted the same balance co-control strategy as described for the standing experiments in the previous chapter (Figure **??**). However, lateral pelvis perturbation was used in the walking trials which were followed by a participants' left or right hip abduction and side stepping. 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. Please, refer to subsection 1.2 for details on the balance assistance controller and perturbation detection.

2.3 Experiment

2.3.1 Experimental setup and balance assistance.

The LOPES lower limb robotic exoskeleton which allows active force control in knee flexion/extenstion, hip extension/flexion, hip ab-/adduction and horizontal pelvis movements

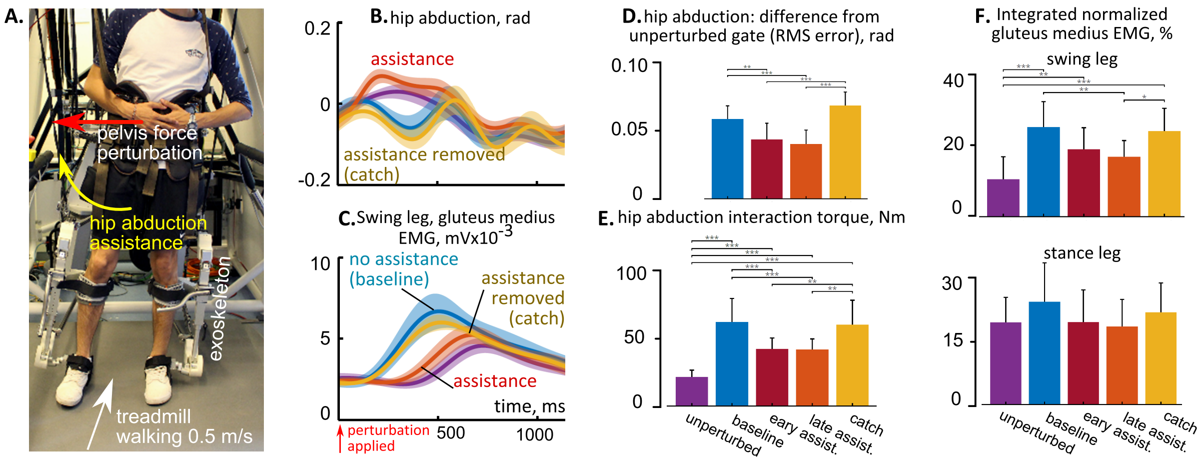

Figure 2.1: **A:** Experimental setup. **B-E:** mean single subject results for different experimental phases. **B:** Hip abduction after the perturbation. **C:** Swing leg gluteus medius EMG activation after the perturbation. **D:** Difference of hip abduction after the perturbation from non-perturbed walking. **E:** Hip abduction interaction torque. **F:** normalised integrated gluteus medius activation (averaged for all subjects) for swing (top) and stance (bottom) legs after the perturbation.

was used in the experiments [22]. Seven healthy adult subjects took part in the study which was conducted at University of Twente and was approved by the institutional ethical committee. The participants were required to walk (0.5 m/s) on the treadmill with the exoskeleton attached to the lower limbs and pelvis (Fig. 2.1A). LOPES actuators were operated in haptically transparent mode, with the pelvis lateral active DoF used for force perturbations (30-40% of a participant's body weight) and left/right hip abduction DoF used for providing assistance torque to support stepping in balance recovery after the perturbation. The exoskeleton balance recovery assistance was composed of short but relatively large decaying in time assistive torque pulses for hip abduction once the lateral body perturbation was detected (when pelvis lateral acceleration and jerk exceeded the predefined levels).

2.3.2 Protocol.

The experiment was composed of four phases: 1) familiarisation, 2) baseline: walking with perturbations in transparent exoskeleton, 3) assistance: walking with perturbation with robotic assistance, 4) catch trials: walking with perturbations with assistance removed randomly in 20% of the cases. In the first phase the participants walked in the exoskeleton under transparent mode control for about 3-4 minutes to familiarise themselves with the machine. Lateral pelvis force perturbations were introduced in the Baseline stage (20 left and 20 right shuffled and randomly applied every 2nd to 4th gait cycle). The perturbations were applied such that a participant was constrained to use the swing leg for lateral stepping during balance recovery (for example, perturbation to the left was applied when the gait transitioned from double stance to left leg swing and right leg stance). While no robotic assistance was introduced during the baseline phase, the hip abduction assistance was introduced every time the perturbation was detected in the assistance phase of the experiment. The goal of the assistance phase was to investigate the effects of robot assistance to balance recovery and to evaluate the quality of cooperative human-robot actions. The assistance phase contained 35 left and 35 right perturbations followed by hip assistance.

The last phase of the experiment was similar to the assistance case, but randomly selected 10 of the 40 perturbations were not followed by the hip assistance. We introduced these no-assistance 'catch' trials in order to evaluate whether there was any learning in the assistance phase.

2.3.3 Data recording and processing.

Exoskeleton's hip/knee joint flexion and abduction angles, pelvis position were recorded at 1 kHz. Participant's EMG data from the left and right leg gluteus medius muscles were recorded at 1 kHz and processed (de-meaned, high-pass filtered at 250 Hz, rectified, low pass filtered at 10 Hz and normalised per subject per muscle). The normalised EMG signal was then integrated over the average duration of the swing and stance phases. To analyse the changes in kinematics of leg movement we calculated the root mean square deviation of hip abduction of the swing leg during balance recovery action with simple unperturbed walking. For within-subject analysis, a one-way ANOVA was run for each observation, followed by post-hoc t-tests with Tukey correction. In the analysis, the assistance phase was split into early assistance (first 20 trials) and late assistance (last 20 trials) for the remaining trials.

2.4 Results

Fig. 2.1B-C shows the hip abduction and muscle activation for a typical subject over all trials per experimental phase, respectively. Both hip trajectories and swing leg muscle activation with robotic assistance were different from the baseline recordings, and the catch trials' recordings were similar to the ones of the baseline stage. Next we present the within-subject statistical analysis of major observations.

We evaluated the difference of perturbed walking (w/ and w/o assistance) to unperturbed walking by comparing the root mean square deviation from unperturbed walking. When assistance was added, hip trajectory was closer to the one recorded during unperturbed walking when compared to the hip abduction without robotic assistance. As shown in Fig. 2.1D the difference with unperturbed walking for early and late assistance phases is significantly lower than Baseline ($p = 0.002$, for 5/7 subjects; $p = 0.0002$, for 5/7 subjects, respectively) and Catch phases ($p < 0.001$ for 6/7 subjects), as seen in Fig. 2.1D. This indicates that the assistance resulted in kinematics more similar to the unperturbed walking than occurred in unassisted perturbed walking. Additionally, assistance caused the swing leg to move out earlier than when no assistance was applied.

As shown in in Fig. 2.1E, there was a reduction in peak torque when assistance was added, from Baseline to Early Assistance ($p < 0.05$ for 5/7 subjects) and Baseline to Late

Assistance ($p < 0.05$ for 6/7 subjects). When the assistance was removed in Catch trials, interaction torque was higher than Early Assistance ($p = 0.003$ for 5/7 subjects) and Late Assistance ($p = 0.002$, 6/7 subjects).

Fig. 2.1E shows recorded integrated and normalised EMG of swing leg (top plot) and stance leg (bottom plot) across all participants. There was a reduction in EMG from Baseline to Late Assistance ($p < 0.05$ for 7/7 subjects) which was also lower compare to Catch. Muscular contraction was consistently higher in all perturbed conditions than in unperturbed walking ($p < 0.006$ for 6/7 subjects) except Late Assistance, which did not show significance in between-subjects analysis ($p = 0.07$ for 6/7 subjects). Between subjects analysis of gluteus medius EMG of the leg in stance during balance recovery showed no significant increase or decrease for assisted and non-assisted cases. The reduction in swing leg EMG from Baseline to Late Assistance shows that the applied assistance reduced the muscular effort required of subjects to successfully recover balance. No difference between Early and Late Assistance implies that there was not a major learned component to this reduction. Similarly, no difference between Baseline and Catch suggests no difference in the recovery when subjects expected assistance.

2.5 Discussion

The results of the experimental study suggest that simple control approaches are sufficient to provide assistance to lower limb movements during balance recovery. All participants were able to easily adapt to robotic assistance. Exoskeleton provided assistance which did not lead to unnatural and less-safe gaits. Muscle activation measurements demonstrated that our approach to assistance can reduce EMG and the participants were able to reduce their muscular effort to achieve their goal. Same level of the muscle activation in baseline and catch trials suggests that no learning was required to adjust the participants' motor control.

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