The Effects on Kinematics and Muscle Activity of Walking in a Robotic Gait Trainer During Zero-Force Control

Edwin H. F. van Asseldonk, Jan F. Veneman, Ralf Ekkelenkamp, Jaap H. Buurke, Frans C. T. van der Helm, and Herman van der Kooij

***Abstract*—“Assist as needed” control algorithms promote activity of patients during robotic gait training. Implementing these requires a free walking mode of a device, as unassisted motions should not be hindered. The goal of this study was to assess the normality of walking in the free walking mode of the LOPES gait trainer, an 8 degrees-of-freedom lightweight impedance controlled exoskeleton. Kinematics, gait parameters and muscle activity of walking in a free walking mode in the device were compared with those of walking freely on a treadmill. Average values and variability of the spatio-temporal gait variables showed no or small (relative to cycle-to-cycle variability) changes and the kinematics showed a significant and relevant decrease in knee angle range only. Muscles involved in push off showed a small decrease, whereas muscles involved in acceleration and deceleration of the swing leg showed an increase of their activity. Timing of the activity was mainly unaffected. Most of the observed differences could be ascribed to the inertia of the exoskeleton. Overall, walking with the LOPES resembled free walking, although this required several adaptations in muscle activity. These adaptations are such that we expect that Assist as Needed training can be implemented in LOPES.**

***Index Terms*—Electromyography, neurorehabilitation, robotic gait trainer, stroke.**

I. Introduction

Development of robotic devices for gait rehabilitation of stroke patients is motivated by the need for intensive training, which was shown to be the key element in facilitating recovery [1]–[3], together with the need for a therapist-friendly training. Now that several robotic devices for gait rehabilitation, i.e., Lokomat [4] and Gait Trainer [5], are available on the market [6], most research effort is put in determining the effectiveness of these devices [7], [8] and in introducing new principles and concepts to extend the possibilities compared to this “first generation” of gait training devices [9]–[12]. These “first generation” devices can be characterized by the approach of enforcing gait upon a patient by moving the legs through a prescribed gait pattern. Although this approach has been proven to be effective in retraining severely affected patients [7], [8] , treatment outcome might be optimized by increasing the active participation of the patient.

Active participation may be realized by increasing motivation [13] or by necessitating self generated activity. The latter can be achieved by adjusting the robotic assistance to the actual abilities and actions of the patient, so that the patient only receives “assist as needed” and performs subtasks of walking where possible on own effort. The potential of an assist as needed (AAN) algorithm in promoting recovery, has been shown by Cai *et al.* [14], who taught spinal mice to step with different robotic control strategies. The mice trained with an AAN algorithm showed a larger recovery of stepping ability than mice trained with a fixed trajectory. These results are not yet confirmed in gait training of human subjects. However, implementation of an AAN algorithm in arm training using the MIT-Manus showed a larger decrease in impairments in the trained stroke patients compared to the strategy of active assistance [15]–[17].

In gait training “assist as needed” can be implemented by only assisting affected subtasks of walking and leaving other subtasks to the patient, for example only assisting the affected leg in case of hemiplegia. This resembles what a therapist is commonly doing during gait training. Position controlled devices are not suited to implement such interventions as they largely enforce motion patterns. AAN strategies require interaction control, usually called haptic- or impedance-control [12]. By adjusting its impedance, the behaviour of a robot can be varied from very stiff to very flexible. A very stiff control mode would then resemble a position control. Next to this, on the other side of the stiffness spectrum, a “zero-impedance” or “zero-force” mode should be available. Here, the subject is able to move freely with minimal resistance of the robot. Any possible intervention during training lies in the large range of possibilities in-between these both extremes. In our project we called these extreme and opposing modes “robot-in-charge” and “patient-in-charge” mode of the robot, respectively.

The availability of a “patient-in-charge”-mode is important for two reasons. First, it is the basis for any “assist as needed” control or selective control [18] algorithm. Only assisting

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affected functions or subtasks implies that the remaining gait functions should be performed unhindered, or in other words 'no assistance when not needed'. Second, walking in the "patient-in-charge" mode can be considered the final stage of training, the function the patient is training towards. The better this mode functions, the more the self-initiated walking with the device will resemble walking without the device and the more likely it is that the acquired capabilities will transfer to over ground walking. For instance, when the device resists the amount of movement in a joint, subjects need to generate more activity to overcome this resistance. However, when walking without the device this activity will lead to unwanted exaggerated movements. The requirement to resemble normal walking also fits in the philosophy that motor training should be task-specific [19]–[21], meaning that training should consist of a meaningful task and that the task in training should strongly resemble the task that has to be learned.

How well a patient-in-charge-mode can be implemented, largely depends on the mechatronical design of the used device. Not only the quality of impedance control on all actuated joints is important, but also *which* of the natural human motions [or degrees of freedom (DoFs)] are actuated, left free, or restrained in the device. The DoFs of the device determine which motions occurring during normal or impaired free walking will be restrained by the device, and which will be available. Evaluation of the "patient in charge" mode reveals how the device affects the different aspects of natural gait. This information is required for a better understanding of the patient's response during training in the device.

No impedance controlled robotic gait training device has yet been evaluated in a patient-in-charge mode. Evaluation of the Lokomat, the device used in most research in the field, has only profoundly been carried out in the position control mode [22]–[25]. The Lokomat can also function in a low impedance mode, but only preliminary results have been published [10]. As the Lokomat was originally not designed to function as an impedance controlled device, we developed and built a light weight device called LOPES (LOwer extremity Powered ExoSkeleton) [26], [27]. Apart of being impedance controlled, it differs from the Lokomat in that it has more actuated degrees of freedom. Besides the common hip and knee flexion and extension, the LOPES allows pelvis translations in the horizontal plane, and hip ab-/adduction, These additional DoFs may be beneficial for training as they allow to leave balance control related tasks to a patient [25]. The importance of adding pelvic motions to gait training, has recently also been pointed out by Aoyagi and Colleagues [12] in the design of the PAM gait training device, which specifically focuses on supporting pelvic motions, including rotations, during training. The addition of these DoFs adds an extra dimension to the Assist As Needed algorithms, as the added DoFs can again be blocked by control, but can be added by choice of the therapist in control of the robot when the walking capacity of the patient allows.

The goal of the present study is to evaluate the "patient-in-charge" mode in LOPES, by comparing walking with and without the device. The criteria for evaluation are first of all how close the kinematics and basic gait parameters while walking with the device resemble those in free walking. However, not only the kinematics but also the muscle activity underlying these walking movements should be similar. We quantified the amplitude of the muscle activity as this provides an indication of the subject's response to the experienced resistance or restraints. In addition we quantified the timing of the activity, to get a measure of the muscle coordination. Finally, as the neural control of walking is characterized by cycle-to-cycle variation, we also assessed whether walking in the device showed the natural variability. The results can be used to deduce possible aspects for further improvement of the design of the robot. Furthermore, the results provide reference values for training with patients, as no improvements have to be expected beyond the observed deviations of normal walking in healthy subjects.

II. Materials and Methods

A. Subjects

Ten healthy young adults (four male, six female, weight 74.9 ± 16.3 kg, height 1.81 ± 0.10 m), mean age 25.9 years volunteered to be participants for this experiment. All participants provided informed consent before testing began.

B. Experimental Apparatus and Recordings

1) Rehabilitation Device: For the experiments the prototype of the gait rehabilitation robot LOPES was used. LOPES is an exoskeleton type rehabilitation robot. The exoskeleton was designed to have low weight and inertia, which was achieved by placing the actuators away from the moving frame, using flexible Bowden cables to transfer the power from the fixed actuators to the freely moving joints of the exoskeleton [Fig. 1(b)]. The inertial characteristics of the robot are presented in Table I. These are relevant because in the current version of the device we are not able to compensate for the inertia of the robotic limbs.

The exoskeleton is actuated by Bowden cable driven series elastic actuators [28]. It offers 3-D translations of the pelvis, where the antero–posterior and medio–lateral motions [1 and 2 in Fig. 1(a)] are actuated. Furthermore, it has two actuated rotation axes in the hip joints [4 and 5 in Fig. 1(a)], and actuated knees with one rotation axis [6 in Fig. 1(a)] [26]. The robot is impedance controlled, which implies that the actuators are used as force (torque) sources. This allows implementing both the "robot-in-charge" mode, that is position control with high impedance, and a "patient-in-charge" mode, that is zero-impedance, in this case synonymous with zero-force control, on each DoF [28]. In the latter mode LOPES was controlled to provide minimal resistance during walking. This was implemented by a closed-loop force controller that controlled torques at joint-level to zero. The "patient-in-charge mode" allowed us to investigate to what degree "free" walking with the device resembled free walking on a treadmill.

2) Motion Capturing: Motions were measured with an PTI Phoenix Visualeyez VZ4000 system (PTI Phoenix, Burnaby, BC, Cananda) at a frequency of 60 Hz. Twenty five uniquely identifiable infrared markers were attached to track the motion of the subject's left leg and exoskeleton's left "leg." Motion of the upper leg, lower leg and trunk was measured by attaching frames with four infrared markers on the back of the thigh, on

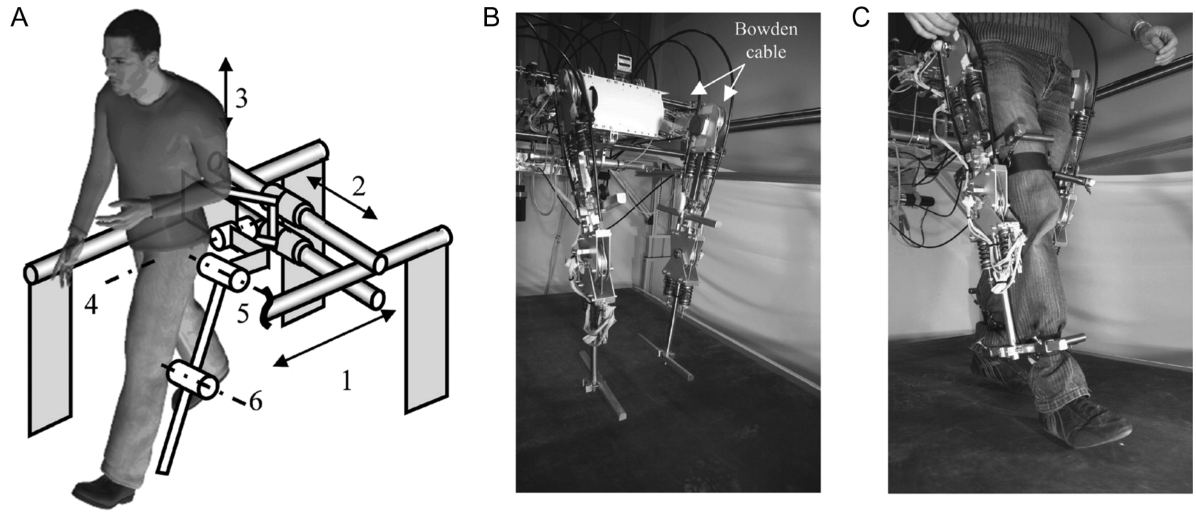

Fig. 1. A) Schematic overview of the LOPES exoskeleton and its DoF. The horizontal movements of the pelvis (1 and 2) are actuated, as well as the movements of the leg joints (4, 5, and 6). Vertical movement of the pelvis is unactuated. Three degrees of the not depicted left exoskeleton leg complete the number of DoFs to nine. B) Exoskeleton with its bowden cables to transmit the forces from the motor to the joint. C) Fixation of the subject in the LOPES exoskeleton.

TABLE I
APPROXIMATE INERTIAL AND MASS CHARACTERISTICS OF THE ROBOT FRAME. FOR REFERENCE APPROXIMATE VALUES FOR A HUMAN PERSON WITH A BODY MASS OF 75 kg ARE GIVEN. FOR TRANSLATIONAL DOFS NO VALUES FOR THE INERTIA ARE PROVIDED

Segment	Involved DoF	Robot		Corresponding human body segments	
		Mass	Inertia	Mass	Inertia
Total robot	Forward/ backward	35	-	75 *	-
	Sideways	27	-	75 *	-
Upper leg	Hip Flex/Ext	2.9	.088	9.5	.15
Lower leg	Knee Flex/Ext	2.2	.064	3.5	.06

* 45 kg only trunk

Mass in kg and inertia in kgm^2

the back of the shank and to the vertebra prominence. The foot motion was measured by placing markers on the ankle, heel, fifth metatarsophalangeal joint, and dorsum. The motions of the exoskeleton were measured by attaching four markers on the upper leg and lower leg part of the exoskeleton.

Gait phases were detected with footswitches, taped directly to the subject's heel and fore foot of both feet. The measurement of the footswitches was synchronized with the measurement of the EMGs.

3) Muscle Activity Measurement: During all trials muscle activation patterns were determined by recording bi-polar surface electromyography (EMG) from the gastrocnemius medialis, tibialis anterior, biceps femoris, rectus femoris, adductor longus, vastus lateralis, and gluteus medius and maximus muscles of the right leg. Skin preparation and the placement of the disc-shaped solid-gel Ag/AgCl-electrodes (type H93SG, Tyco Healthcare/Kendall, Mansfield, MA) in a bipolar configuration were performed according to Seniam guidelines [29]. For the EMG recordings a compact measurement apparatus (type Porti 16-5, TMS International, Enschede, The Netherlands) was used. The analog signals were sampled at 1024 Hz and sent from the portable unit via fibre optics to the computer, where data were stored for further processing.

C. Experimental Protocol

The subjects walked freely on the treadmill and while strapped into LOPES, the order of the "type of walking" was randomized between subjects. For each type of walking, the subject walked at 0.5, 0.75, and 1.25 m/s. Within each type of walking the order of the velocities was randomized. With each change of walking velocity and/or type of walking the subject was given 3 min to get used to the walking condition. During this period and subsequent testing, subject did not receive any specific instructions about how to walk in the device. Subsequently, data were recorded for approximately 20 steps in total.

Before the subjects could walk with LOPES, the exoskeleton was attached to the subject's leg and pelvis [Fig. 1(b)]. The subject's joint axes were aligned with the joint axes of the exoskeleton by adjusting the pelvis width of the exoskeleton and the length of exoskeleton linkages. As the exoskeleton did not encompass an ankle joint, the ankle was left free to move.

D. Data Analysis

1) Joint and Segment Kinematics: The kinematical data were split into individual stride cycles. The individual marker paths were filtered by the rigid body filter of the PTI software (VZ-Analyzer V3.50). This filter allows grouping sets of markers in supposed rigid bodies, and uses the motion paths of all markers of the body to reconstruct and filter the motions of the separate markers, mainly based on the impossibility of shape changes and marker jumps. The chosen rigid bodies were: trunk cluster, upper leg cluster, lower leg cluster, foot, robot upper leg, and robot lower leg. All positions were expressed in a coordinate system defined by the walking direction (x), the vertical (y), and the axis perpendicular to this plane (z) according to the right hand orientation.

For calculating segment angles, the projections of the orientation vector of the marker-clusters on two global planes (sagittal-xy and frontal-yz) were used. We defined the straight standing posture as zero angle for the different segments. The knee angle was obtained by subtracting the lower leg angle from the upper leg angle. As pelvis rotations during walking are relatively small compared to the thigh rotations we assumed that the thigh angles reflect the rotations in the hip. To illustrate the average trajectories for each condition, the angles for each cycle were time normalized and subsequently averaged over all cycles and ensemble averaged over all subjects.

From the calculated angles, we extracted for each step-cycle the angular range of motion of the frontal and sagittal trunk and thigh angle and the sagittal knee angle.

2) Gait Parameters: We also compared the spatial and temporal gait parameters [30], [31] We calculated the step width from the measured marker positions. As only the movements of the left leg were measured, we defined the step width as twice the z distance between the average position of the left ankle marker during single stance for each cycle and the whole trial average trunk position, obtained from the four trunk markers.

The footswitch data were used to calculate the temporal variables including cycle time (period between two consecutive contacts of the same foot), stance time (period between initial contact and toe-off in the same limb), swing time (period from toe-off to heel contact of the same limb) and double stance ratio (total double stance time divided by the single stance time). Other gait parameters such as step length and double stance time were also determined but as they are linearly related to afore described parameters, they were not used in statistical testing to limit the number of comparisons.

3) Within Subject Variability: To investigate the within subject variability, we assessed the standard deviation over the values for the different cycles for each separate condition and subject for the aforementioned spatial and temporal gait parameters and for the frontal and sagittal angular ranges of motion.

4) EMG Measurements: All EMG processing was done with custom written software [32] in Matlab (Natick, MA). The raw EMG data were band pass filtered at 10–400 Hz with a second-order zero-lag Butterworth filter and converted to smooth rectified EMG signals (SRE) using a low-pass second-order zero-lag Butterworth filter at 25 Hz for smoothing. To visually inspect the raw and smooth rectified signals, they were broken up into the individual stride cycles. If one of the muscles contained artifacts (contact artifact, measurement noise) the activity during this cycle was rejected from further analysis. Subsequently, the SRE of each muscle was normalized to its maximal activity over the entire experiment. The mean EMG was calculated from these SRE traces over seven intervals of walking. The values for all the gait cycles for a single condition were averaged to result in one value for each combination of subject, muscle, interval, velocity, and type of walking.

The different intervals of walking were defined based on the footswitch data of both feet (see Fig. 2). The double stance phase starting with heel contact of the left leg was defined as initial loading and the second double stance phase was defined as pre swing. Although loading and swing preparation were not the only ongoing process during the different double stance phases, we decided to call the double stance phases after these processes as these processes were the most dominant ongoing processes during these phases. The single stance phase was split up into two intervals of equal length: midstance and terminal stance and the swing phase was split up into three intervals of equal length: initial swing, mid swing, and terminal swing.

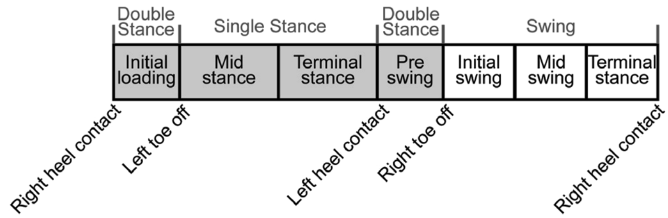

Fig. 2. Segmentation of the gait cycle into intervals for the calculation of the integrated EMG over these intervals. For clarity, the standard gait events are indicated with respect to these intervals.

The smoothed EMG data were used to determine the onset and cessation times of the main burst for each muscle. The used algorithm [32] is based on the approximated generalized likelihood ratio (AGLR) principle described by Staude and Wolf [33]. This algorithm was used to analyze the smoothed EMG signals of every stride in the gait cycle separately with respect to the on- and off-times of muscle activation. Subsequently, all detected on- and off-times were normalized in time using the stride time starting from the related heel strike.

E. Statistical Analysis

Linear mixed modeling analyses were applied to explain differences in mean EMG activity over time by the factors type of walking (two-level factor "type of walking") for each velocity (three-level factor "velocity") per interval (seven-level factor "interval"), separately for each muscle.

To account for the correlation between the repeated measurements within a subject, different intercepts were assumed for each interval per subject (by including the factor subject and interval as random factors). The factors "type of walking," "velocity" and "interval" were treated as fixed effects. The two-way interactions "type of $\mathrm{walking} * \mathrm{velocity}$," "type of $\mathrm{walking} *$ $\mathrm{interval}$," and '$\mathrm{velocity} * \mathrm{interval}$" and three-way interaction "type of $\mathrm{walking} * \mathrm{velocity} * \mathrm{interval}$" were also included. For all significant effects and interactions *post hoc* tests (Sidak adjustment) were performed. The level of significance was defined as 5%.

For the onset and cessation times of EMG bursts, kinematics and gait parameters, a similar mixed model was used without the interval factor, with subject as a random factor, and type of walking and velocity as fixed factors.

III. Results

The effect of walking with LOPES was assessed by comparing temporal and spatial gait parameters, movements and muscle activity between LOPES walking and free walking. The results on each of these aspects are presented in the following paragraphs.

A. Gait Parameters and Kinematics

The basic gait parameters showed some significant changes between walking in LOPES and treadmill walking (see Table II). The cycle time and total stance time did not show a significant change, whereas the swing time showed a significant increase. The increase in swing time was accompanied

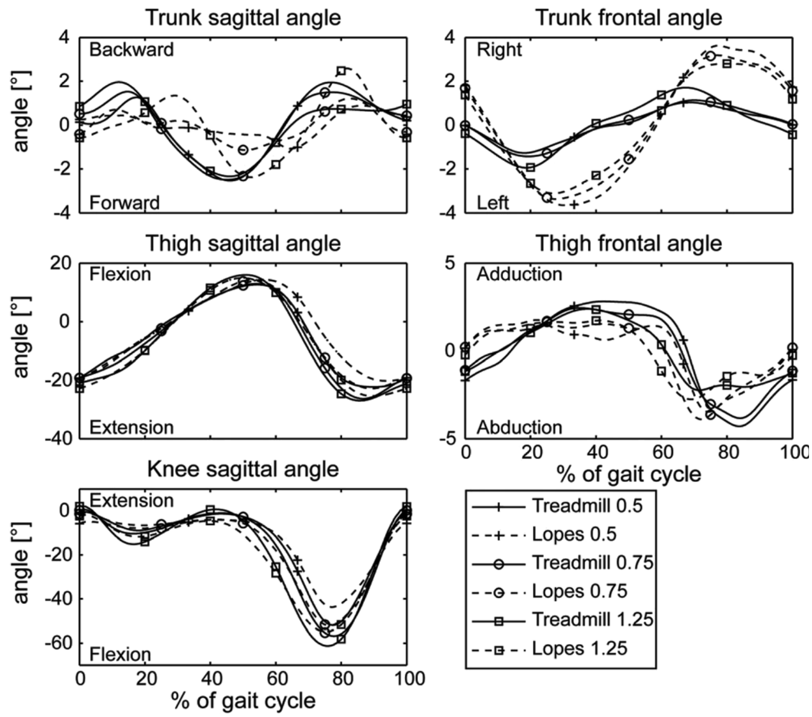

Fig. 3. Average trajectories of the sagittal and frontal rotations of the trunk (upper panels), of the upper leg (middle panels), and the sagittal angle of the knee (lower panel) for all conditions, averaged over all subjects.

TABLE II
AVERAGE VALUES AND WITHIN-SUBJECT VARIABILITY OF THE GAIT PARAMETERS (± STANDARD DEVIATIONS) FOR THE DIFFERENT TYPES OF WALKING

Gait parameter	Average		Within subject variability	
	Treadmill walking	LOPES walking	Treadmill walking	LOPES walking
Cycle time	1.43 ± .31 s	1.46 ± .33 s	.031 ± .020 s	.034 ± .023 s
Double stance ratio	.45 ± .05	.40 ± .06 *	.028 ± .017	.033 ± .020
Swing time	.47 ± .08 s	.51 ± .09 s*	.032 ± .015 s	.035 ± .027 s
Stance time	.96 ± .23 s	.96 ± .25 s	.018 ± .008 s	.018 ± .011 s
Step width	.33 ± .06 m	.34 ± .05 m*	.023 ± .006	.027 ± .010 m*

* significantly different from treadmill walking with p<.05

TABLE III
AVERAGE AND WITHIN-SUBJECT VARIABILITY OF THE ANGLE RANGE-OF-MOTION (± STANDARD DEVIATIONS) FOR THE DIFFERENT TYPES OF WALKING

Angle range	Average		Within subject variability	
	Treadmill walking	LOPES walking	Treadmill walking	LOPES walking
Sagittal trunk	5.7 ± 1.2	5.6 ± 1.8	1.2 ± .4	1.4 ± .5*
Frontal trunk	4.9 ± 1.9	7.8 ± 2.3 *	1.2 ± .5	1.4 ± .4
Sagittal thigh	42.9 ± 10.8	38.8 ± 9.7 *	2.5 ± .9	2.0 ± .8*
Frontal thigh	8.8 ± 3.3	8.0 ± 3.0	0.9 ± .4	1.0 ± .4
Sagittal knee	66.0 ± 11.0	54.1 ± 8.4 *	3.3 ± 1.5	2.9 ± .8

* significantly different from treadmill walking with $p < 0.05$

by an increase in single stance time, which led to a significant decrease in Double stance ratio ($p < 0.001$). The step width showed a small (0.014 m) but significant ($p = 0.011$) increase in LOPES walking compared to treadmill walking.

The changes in gait parameters were accompanied by changes of the segment/joint kinematics (Fig. 3). The sagittal knee and thigh rotations and the frontal thigh rotation showed similar patterns in both conditions. Still, the range of motion for the knee ($p < 0.001$) and sagittal ($p = 0.005$) thigh movements were significantly smaller in LOPES walking (see Table III). The general trajectories of the trunk diverge considerably more between the different modes of walking. For the frontal trunk rotation, this was accompanied by a significant ($p < 0.001$) increase of the range of motion during LOPES walking.

B. Within Subject Variability

In order to assess whether walking in LOPES influenced the amount of within subject variability, the standard deviation of the values of the different cycles were calculated for each subject and condition and compared between conditions. The within subject variability for the cycle time, stance time, swing time, double stance ratio, and angular ranges of the knee and frontal trunk and thigh did not show any significant difference (see Tables II and III). The variability in the step width ($p = 0.032$) and the sagittal trunk angle range ($p = 0.015$) were significantly larger in LOPES walking, whereas the variability in the sagittal thigh angle was significantly smaller ($p = 0.049$). Still, the absolute differences were small, 0.004 m, 0.2° and 0.5° respectively. None of the gait parameters or kinematic parameters showed an interaction effect between type of walking and velocity.

C. Muscle Activity

An evaluation of the movements does not suffice in determining whether unhindered walking is possible in the LOPES device, as subjects could have produced the same movements at the cost of different muscle activity in coping with the device. Therefore we also evaluated the muscle activity in timing and in amplitude.

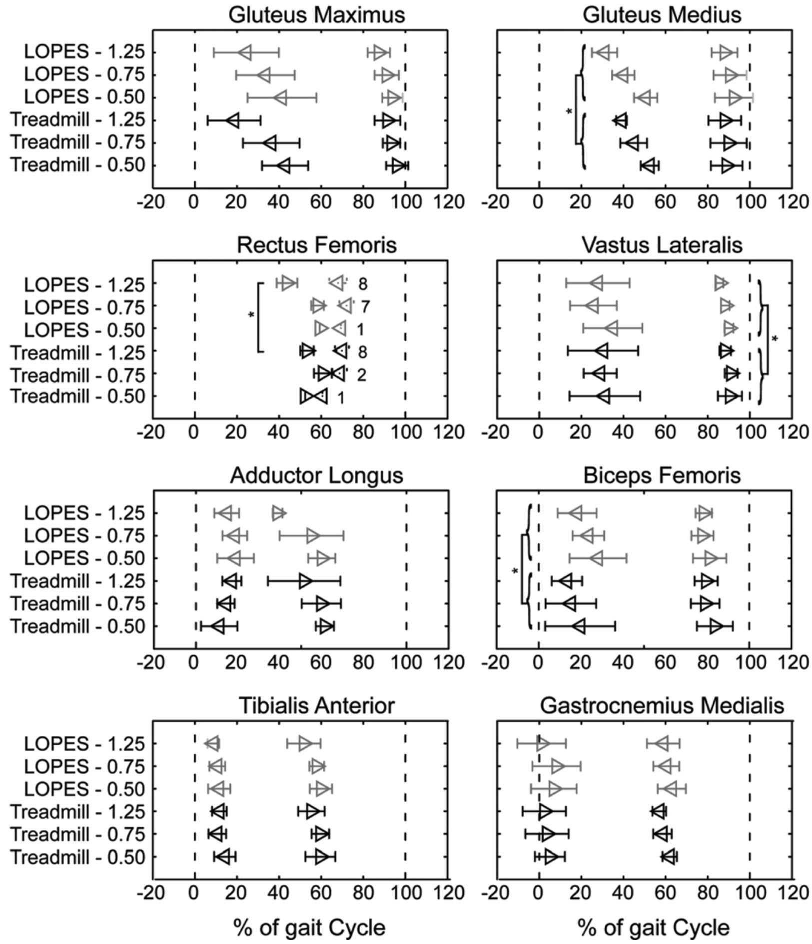

Fig. 4. Mean and standard deviation of the onset and cessation time of the main burst for each muscle. On the left side the conditions are indicated. Grey symbols indicate the different LOPES conditions and the black symbols the different treadmill conditions. The ▶ symbol indicates the onset of the burst and the ◀symbol indicates the cessation of a burst. The numbers in the rectus femoris panel specify the number of subjects who showed a burst in that condition. Significant differences are indicated with the symbol *.

1) Onset and Cessation Times of Burst: The onset and cessation times of the main bursts reflect the timing of muscle activity (see Fig. 4). For the rectus femoris, we only used the burst during stance-to-swing transition, as this burst reflects the activity of the rectus femoris and is not caused by cross talk from the vastus lateralis and vastus intermedius as is the case with the often reported burst during initial contact [34]. The occurrence of the burst in the rectus femoris was strongly dependent on the walking velocity and the type of walking. For the intermediate velocity (0.75 m/s) only two subjects showed a burst during treadmill walking whereas seven subjects showed a burst during LOPES walking (see numbers in the rectus femoris panel of Fig. 4). The inequality in the number of subjects showing the burst could be considered as a measure of the difference in treadmill and LOPES walking. For the highest velocity, only two subjects did not show a burst, the remaining subjects showed a significant earlier onset in LOPES than in treadmill walking (43.8% versus 53.3% walking, $p < 0.001$), whereas the cessation did not change significantly. For the other muscles, the occurrence of a burst was neither dependent on the type of walking nor on the walking velocity. The timing of the burst changed for three other muscles. The onset occurred significantly earlier for the vastus lateralis in LOPES walking compared to treadmill walking (88.0% versus 90.0%, $p = 0.031$). The burst of the gluteus medius ended significantly earlier in LOPES walking compared to treadmill walking (40.5% versus 45.4%, $p < 0.001$) whereas the biceps femoris showed a delayed cessation while walking in LOPES (16.1% versus 23.4%, $p = 0.024$). None of the muscles showed an interaction effect between velocity and condition.

2) Mean Integrated Activity: Although all muscles showed approximately the same pattern during LOPES walking and treadmill walking (Fig. 5), they all showed a significant difference in the integrated activity for at least one interval. The hip extensors (gluteus medius and gluteus maximus) showed a decrease of their activity while walking in LOPES (significant type of walking effect, see Table IV). This difference was dependent on the interval. During the start and the end of the stance phase, the mean activity was significantly lower, whereas in the other intervals there was no significant difference. For the knee extensors (vastus lateralis and rectus femoris) the overall activity did not differ significantly. Still, for the rectus femoris the activity was higher during the transition from stance to swing. In addition, during initial loading the activity of both knee extensors was lower in LOPES. However, the activity of the rectus femoris during initial loading was most likely the

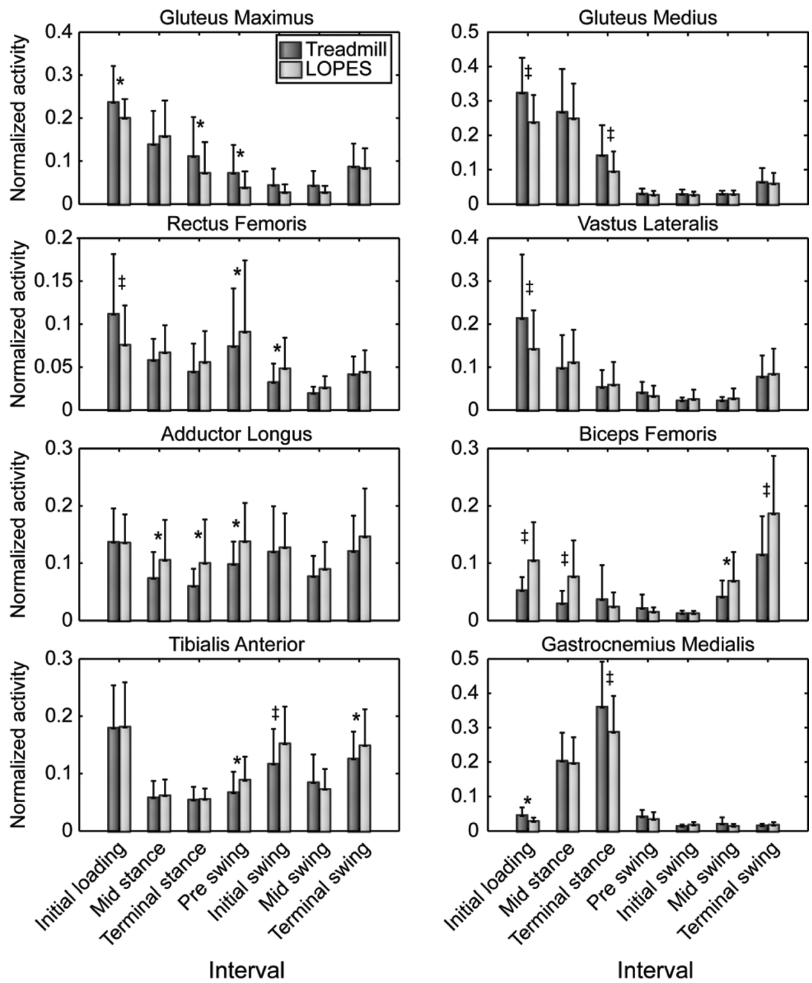

Fig. 5. Mean normalized integrated activity for each muscle over the different intervals for LOPES walking and treadmill walking. Vertical bars indicate the standard deviation over the different subjects. Significant difference between LOPES walking and treadmill walking are indicated with a * for $p < 0.05$ and with a ‡ for $p < 0.001$.

TABLE IV
SIGNIFICANCE LEVELS FOR THE MAIN EFFECT OF TYPE OF WALKING AND ALL INTERACTION EFFECTS CONTAINING TYPE OF WALKING FOR THE COMPARISON OF THE MEAN NORMALIZED EMG ACTIVITY

Muscle	TW	TW x I	TW x V	TW x V x I
Gluteus maximus	<.001	.012		
Gluteus medius	<.001	<.001		
Rectus femoris		<.001		
Vastus lateralis		<.001		.001
Tibialis anterior	.001	.001		
Biceps femoris	<.001	<.001		
Adductor longus	<.001		.006	
Gastrocnemius medialis	<.001	<.001		

TW indicates the factor Type of walking, I indicates the factor Interval, V indicates the factor Velocity

consequence of cross talk from the vastus lateralis and vastus intermedius [34]. The significant third-order interaction for the vastus lateralis indicated that the difference in type of walking during initial loading was dependent on velocity. For the middle and high velocity, the difference was significant.

The bi-articular biceps femoris showed an increase of its activity for LOPES walking. This difference was dependent on the interval, only the intervals from mid swing to mid stance were higher. Exact during these intervals the main burst of the biceps femoris occurred. The adductor longus was the only muscle that showed an interaction between type of walking and velocity. For the highest velocity the activity was significantly higher in LOPES walking, whereas for the middle and lowest velocity the difference was not significant. Both lower leg muscles showed a type of walking effect and an interaction effect between type of walking and interval. They showed their most prominent differences during the transition from stance to swing. The gastrocnemius had a smaller activity during terminal stance in LOPES walking than in treadmill walking, whereas the tibialis anterior showed an increased activity during pre swing and initial swing while walking in LOPES.

IV. DISCUSSION

In this study we evaluated the effect of walking in a robotic gait training device (LOPES) in a "zero-force" mode on the kinematics and muscle activity. This "zero-force" mode is of great importance for the implementation of interactive gait training according to the principles of "assist as needed." Walking with LOPES differed significantly on several aspects from walking without the device. These differences occurred in the gait parameters, joint/segment movements and the muscle activity. However, except from the knee angle range, the observed differences in the gait parameters and kinematics were generally small. The average difference between the two conditions for those parameters (that is swing time, step width, sagittal thigh angle, see Tables I and II) was smaller or close to the observed cycle-to-cycle variability for these parameters. Furthermore, the differences were small compared to the accuracy of the measurements and calculations [31], [35], [36]. Walking with LOPES did generally not affect the cycle-to-cycle variability. This finding indicates that it is possible to walk with natural variability in LOPES. In essence, the walking pattern with the device was close to the walking pattern without the device. However, to maintain a similar walking pattern in LOPES, considerable changes in the muscle activity were required.

In the next paragraphs, we will discuss the differences in kinematics and muscle activity, identify possible causes and elaborate upon the significance of these findings for implementation of AAN algorithms, before we go into the clinical relevance of the findings.

A. Causes of Observed Differences

1) Pelvis Motion: The amplitude of the muscle activity showed several differences, especially for the biceps femoris and the adductor longus. The increased activity of the adductor longus and the frontal trunk rotation, might have their origin in the resistance of the device to lateral movements of the pelvis. The controller did not compensate for the inertia of the exoskeleton and some remaining friction. Less pelvis motion would theoretically lead to less lateral centre of mass excursion. The increase of activity of the adductor longus during stance may be caused by increased effort needed to still push the total body centre of mass to above the weight bearing stance leg. The decrease in lateral pelvis motion is compensated for by an increased frontal trunk rotation. The need to do this is amplified by the increase in step width, which was probably caused by experienced “danger” of walking narrow, because of constructional parts at the legs. The aforementioned changes might have been aggravated by restraining the pelvic rotation, though the pelvic rotations during normal walking are generally rather small [30] and the fixation of the subject into device allowed some movement, which was not measured. A similar study using the PAM [12], which allows pelvic rotation, could further clarify the importance of pelvic rotations in gait training.

In general, the observed changes in muscle activity during the stance phase while walking in LOPES showed lower EMG activity compared to free walking. The gluteus maximus, gluteus medius and gastrocnemius medialis all showed a decrease in activity during initial loading and terminal stance. The gluteus maximus and gastrocnemius medialis play an important role in forward progression during mid stance and terminal stance, respectively [37]–[39]. The drop of their activity indicates a decrease of push off force. Apparently, the contribution of the muscles to the acceleration of the center of mass (CoM) was decreased, which could be the result of an overall decrease of the fluctuations in the acceleration and deceleration of the CoM through continuously opposing forces like friction (in the linear guide) and the inertia. During the experiments, we indeed observed relatively small pelvis-motion while walking in LOPES, compared to free walking [40]. However, a direct comparison was not possible as the experimental setup did not allow measurement of the pelvis motion in both conditions

2) Inertia of Exeskeleton: The changes in activity of the biceps femoris, the rectus femoris and the tibialis anterior, the decreased knee angle range and increased swing time can be addressed to the increased inertia and mass of the swing leg when wearing the LOPES exoskeleton. The swing phase of walking is normally a largely passive, pendulum like motion, driven by the rectus femoris, decelerated by the biceps femoris and semitendinosus. When the exoskeleton is attached to the leg, not only the mass of the leg has to be decelerated but also the mass of the exoskeleton leg. Table I shows that the mass and inertia of the exoskeleton are considerable in proportion to those of the human segments. These values together with the average angle trajectories (see Fig. 3) were used to get a rough estimate of the hip and knee torques required to accelerate and decelerate the swing leg for the different walking velocities. These calculations showed that only for the higher velocities the required torques were substantially increased (up to approximately 5 Nm) when wearing the exoskeleton and that the required torques were mainly higher for decelerating the leg.

The results of this study are in agreement with a study on the effects of additional mass at the lower leg [41]. Adding 2 kg to the left lower leg resulted in changes in the hip and knee torques that correspond with the changes in the muscle activity found in the current study. Adding mass also caused a decrease of maximum knee flexion during swing and an increase of the ankle dorsiflexion, corresponding to the decrease of the maximal knee angle range and the increase of the tibialis anterior activity during the pre swing and initial swing phase. Finally adding mass was shows to increase swing time [41], [42], also in agreement with the present study. So, the observed changes during the swing phase can be ascribed to the effect of the added mass and inertia. In LOPES, device dynamics cannot be compensated by the controller, and will be perceptible for the user. This stresses the need for light weight design and also indicates a point of further improvement in LOPES.

Some muscles also showed changes in timing of their activity. This indicates that different coordination is required. The observed changes were in some cases so small that it could be questioned whether they were also relevant. Based on results of Perry [30], Buurke [43] argued that a change in timing between conditions should be at least 5% to be relevant. The changes in gluteus medialis and vastus lateralis were smaller than 5%, 4.9%, and 2%, respectively, and consequently considered irrelevant. The rectus femoris showed a significant and relevant earlier onset in LOPES walking (9.5%) and the biceps femoris showed a delayed cessation (7.3%) Both of these changes could be attributed to the increased inertia of the swing leg, when the exoskeleton is attached to it. The earlier onset of the rectus femoris indicated an earlier force build up for the initiation of the swing phase, while the delayed cessation of the biceps femoris could have indicated an increase of time to decelerate the swing leg.

3) Comparison With Other Studies: Other studies also compared the activation patterns while walking with a robotic device with the patterns outside the device [12], [44]. However, most of these studies restricted their analysis to a qualitative comparison and did not perform a detailed quantitative analysis. Only Hidler and Wall [45] performed a similar detailed analysis when they assessed the alterations in muscle activation patterns of walking in the position controlled Lokomat compared to free treadmill walking. The muscle activity in their study not only differed during the main burst but also outside these bursts. Especially, the rectus femoris, adductor longus, biceps femoris, vastus lateralis, and gluteus maximus showed clear increases of muscle activity during periods in which the muscle was normally silent. This would probably have resulted in significant changes in the timing of these muscles, however data about the timing were not provided.

Although these results indicate that the muscle activation patterns were closer to normal while walking in LOPES compared to walking in Lokomat, a direct comparison of both studies is complicated by the difference in DoFs of the devices and the used control strategy. Recently, we have shown [40] that the additional DoFs of LOPES (pelvis translations and hip abduction) had only small effects on the results as presented in the current study, especially on the EMG results. Most clear effect of blocking of these DoFs was increased rotations of the trunk around all axes. The normality of EMG patterns in LOPES walking compared to Lokomat walking, should therefore mainly be contributed to the zero-impedance control that replaced the joint-trajectory (position) control.

It is important to note that the interpretation of EMG patterns is cumbersome in a position controlled device, because the generated activity of the subject will hardly influence the resulting motions; different activation patterns would result in approximately the same motions. One is therefore measuring two aspects at the same time: how well a person is actively walking according to the device trajectories (which could also have been passively followed), and how natural these trajectories are.

B. Implications for Assist as Needed

Impedance controlled devices are especially suited to implement assist-as-needed control algorithms, because the magnitude of the provided assistance can then be progressively reduced depending on the stage of recovery of the patient. This will continuously encourage the patient to participate in the training and to improve further. The amount of support can be adjusted for the complete walking pattern or for specific aspects of walking. We have recently developed and implemented an algorithm for the latter [18], [46], in which selective assistance is provided on several subtasks of walking (like foot clearance, step length, lateral balance control, weight support). The performance on each of the subtasks can be evaluated and regulated separately. This means that subjects will only receive assistance on impaired aspects of walking. To decide which movements should be considered impaired, reference values of healthy subjects are required. This study provides such reference values. An important deviation from normal walking was the decrease of maximal knee flexion during swing. Therefore, we should not expect that patients show a return to normal values when the assistance is progressively decreased to zero.

The results also indicated that there was some resistance in the horizontal pelvis motions. These DoFs in the pelvis are essential in balance control during gait. Therefore, the resistance contributes to stabilizing the body and as such hinders the training of balance control with the device. The stabilization might be beneficial early in training. However, it is an undesirable effect when the subject needs to be encouraged to recover his/her balance control.

C. Clinical Relevance

This study showed that healthy people walking with LOPES show walking patterns closely resembling those of walking without the device. It can be hypothesized that this implies that the device has the potency to deliver task specific training. In a device with too large imperfections (considerable friction or inertia, or constrained movements), a patient would learn to generate activity that leads to completely different motions when walking without the device. As mentioned before, the patient-in-charge mode reflects the end stage of training as (ideally) no assistive forces are applied by the robot that interfere with the patients' self-generated forces. It is however not known how close to normal walking the training should be in order to consider it as sufficiently task-specific. This is an important issue that can be resolved only by realizing actual training sessions with patients.

As LOPES is going to be used to retrain stroke patients, we need to assess whether LOPES also minimally affects the typical walking pattern of stroke patients. Due to their impairments, the walking pattern of stroke patients clearly differs from healthy subjects on several aspects [47]–[49]. For example, a large group of stroke patients shows an increased abduction during the swing phase [50]. This increased abduction is part of a hip circumduction strategy which also involves pelvic rotation and which enables patients with a hyper extended knee at push off, to achieve enough toe clearance during swing. Although the amount of pelvic rotation is limited in the device, by utilizing the hip abduction/adduction degree of the exoskeleton, stroke patients could perform these movements in LOPES and consequently would make the movement in LOPES that they usually make without the device. This is especially important as most stroke patients develop their own functional way of walking during training, which is not necessarily the healthy symmetrical way of walking [43].

D. Conclusion

Evaluating the LOPES in a "patient-in-charge" or "zero-force" mode showed that healthy subjects walking with the device show a walking pattern that resembles a normal walking pattern at the expense of changes in the amplitude of the muscle activity. Cycle-to-cycle variation while walking with the device was comparable to free walking. The changes in muscle activity could be attributed to the inertia of the device. Improvements on this aspect can be realised by compensating for or reducing this inertia.

The results indicate that LOPES, or in general an impedance controlled device with sufficiently low possible impedance and sufficient DoFs, can be applied in a gait training that would fit in a progressive training regime that is task specific and following an "assist-as-needed" approach. Whether the shown results are "natural" enough to offer training that teaches skills transferable to free walking remains a question that can only be answered by clinical research.

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