Robot-Aided Gait Training with LOPES

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Edwin H.F. van Asseldonk and Herman van der Kooij

Abstract

Robot-aided gait training in stroke survivors and spinal cord injury patients has shown inconclusive effects on walking ability. It is widely acknowledged that the control and design of the robotic devices needs to be further optimized to be able to provide training that fits better into modern insights in neural plasticity, motor learning, and motor recovery and in doing so improves its effectiveness. We will go more deeply into the need and scientific background for improvements on active participation, task specificity, and the facilitation of different recovery mechanisms. Subsequently, we will discuss recent advances that have been made in the control and design of robotic devices to improve on these aspects. Hereby, we will focus on the robotic gait training device LOPES that has been developed within our group. We will discuss how its design and control approach should contribute to improvements on all of the aforementioned aspects. The feasibility of the chosen approach is demonstrated by experimental results in healthy subjects and chronic stroke survivors. Future clinical testing has to demonstrate whether the outcome of robot-aided gait training can indeed be improved by increasing its task specificity, by the active contribution of the patient, and by allowing different movement strategies.

Keywords

Impedance-controlled exoskeleton • Assist-as-needed • Recovery and compensation • Task specific • Stroke • Spinal cord injury

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V. Dietz et al. (eds.), *Neurorehabilitation Technology*,
DOI 10.1007/978-1-4471-2277-7_21,

21.1 State-of-the-Art Robot-Aided Gait Training

Robotic gait training devices have been on the market since the start of the millennium. Currently, the mechanized gait trainer (Reha-Stim, Berlin, Germany) [1], the Autoambulator (HealthSouth, USA), and the market-leading Lokomat (Hocoma AG, Volketswil, Switzerland) [2] are commercially available. In addition, different research institutes and companies are developing robotic gait trainers, among which are ALEX (Active LegEXoskeleton) [3], a combination of PAM (Pelvic Assist Manipulator) and POGO (Pneumatically Operated Gait Orthosis) [4], and LOPES (Lower Extremity Powered Exo-Skeleton) [5]. All these devices support the patients during treadmill walking. There are also developments in the design of wearable exoskeletons that can be used as assistive or therapeutic devices. The BLEEX (BerkeLEy EXoskeleton) [6], originally developed for military purposes, has been redesigned into a medical exoskeleton called eLEGS. The HAL (Hybrid Assistive Leg) [7] and ReWalk (Argo Medical Technologies, Israel) are other examples of medical wearable exoskeletons that assist during overground walking.

All these devices widely differ in their design and control. The most distinctive feature regarding the design is the number of assisted, free, or constrained degrees of freedom (DOF). Table 21.1 provides an overview of the DOFs of the aforementioned devices. Notably, all commercially available devices only assist movements in the sagittal plane and constrain all the movements out of the sagittal plane, even though these movements are natural to human gait. Regarding the control of the devices, the most distinctive feature is whether the devices control/enforce positions of the limbs or control the interaction forces between the robot and the limbs. Again, the commercially available robotic gait training stands out, as they are position controlled (the new Lokomat is impedance controlled), whereas the other devices are mostly force-controlled.

The effectiveness of robot-aided gait training has only been assessed in clinical trials using the commercially available gait trainers. The first effect studies showed fairly positive results in that training with these devices was at least as effective as manual training while the physical load on the therapists was reduced [8, 9]. Some studies even showed an increase in the number of subjects that could ambulate independently after receiving robot-aided gait training [10]. However, recently, two large randomized clinical trials, one in chronic stroke survivors [11] and one in subacute stroke survivors [12], demonstrated that walking velocity and endurance improved significantly less after robot-aided gait training compared to conventional training. Subacute stroke survivors improved their walking velocity with 71% after conventional training and only 35% after robot-aided training [12]. These latter studies clearly indicated that robot-aided gait training needs to be further optimized to improve its efficiency. Clinicians, (neuro)scientists, and engineers have put forward different ways to advance robotic gait trainers and make robot-aided gait training better fit in with new insights in neural plasticity, motor learning, and motor recovery. In short, the therapeutic benefit of robot-aided gait training might be increased by making the training more task specific, encouraging the patients to actively participate, and facilitating functional improvement by using recovery as well as compensatory strategies.

Advances on these aspects require changes in the mechanical design of the devices and in the control of these devices. The general shift from position to force control and the addition of DOFs in the research devices aim at improving on one or more of these aspects. The robotic gait training device LOPES was specifically developed to improve on all of these aspects. In the following paragraphs, we will first elaborate more on the need to improve on the different aspects to increase the efficiency of robot-aided gait training. Next, we will shortly discuss what achievements have been made in the field of robotic gait training devices, and we will describe the LOPES device into more detail and introduce its mechanical design and control. We will discuss the results that were obtained with the LOPES device and elaborate on the future perspectives.

Table 21.1 Overview of the major features of the mechanical design and control for different robotic devicesa

		Mechanized gait trainer	Lokomat	Autoambulator	ALEX	PAM/ POGO	LOPES	eLEGS	HAL	ReWalk
Mechanical design										
Type[b]		EE	EX	EX	EX	EX	EX	EX	EX	EX
Supports walking on		TR	TR	TR	TR	TR	TR	OG	OG	OG
Degrees of freedom										
Pelvis	Vertical translation	–	F	F	F	A	F	F	F	F
	Horizontal translation	–	C	C	C	A	A	F	F	F
	Rotations	C/–	C	C	C	A	C	F	F	F
Hip	Flexion/ extension	–	A	A	A	A	A	A	A	A
	Abduction/ adduction	C	C	C	C	F	A	A	F	C
	Exoration/ endoration	C	C	C	C	F	C	F	C	C
Knee	Flexion/ extension	–	A	A	A	A	A	A	A	A
Ankle	Plantar flexion/ dorsiflexion	–	F	F	F	F	F	F	F	F
Foot	Vertical translation	A	–	–	–	–	–	–	–	–
	Forward/ backward translation	A	–	–	–	–	–	–	–	–
Control		Pos	Pos/for	For	For	For	For	Pos/for	For	Pos

The device type is either an exoskeleton (*EX*) or end-effector (*EE*). The device is meant to support gait during treadmill (*TR*) or overground (*OG*) walking. The DOFs are actuated (*A*), free (*F*), or constrained (*C*). A dash (–) indicates that the DOF can be indirectly influenced by the provided assistance at the other DOFs

[a]Every year, several new devices are developed and introduced. This table does not give a complete overview of all existing devices

[b]In a pure "end-effector" robot, the interaction of the robot with the human is limited to the "end-effector" of the extremity, the foot. In exoskeleton-type robots, the robot is attached to the controlled limb at several places, and the robot moves in parallel with the segments of the limb

21.2 Background and Rationale for Advancement in Robot-Aided Training

21.2.1 Task-Specific Training Needed for Transfer of Learned Abilities to Overground Walking

Task specificity of training has been shown to be a crucial factor in facilitating functional improvement [13, 14]. Task specificity in this respect means that the trained task should closely resemble the real-world task that needs to be improved. The larger the resemblance, the larger is the likelihood that improvement during training will generalize to the daily task. The task specificity of training in the currently commercially available robotic gait training devices is questionable. This is mainly due to the fact that DOFs that are used while walking overground are constrained in these devices. Although movements in the constrained DOFs are not possible, subjects can still generate torques in those DOFs. For instance, Neckel and colleagues [15] demonstrated that chronic stroke survivors still generated considerable abduction torques during swing when they were walking in a robotic gait trainer that constrained hip abduction movement. These abduction torques reflected that these stroke survivors actually employed a circumduction strategy, but the device was constraining this strategy. When subjects generate the same activity while walking overground, this will result in a hip abduction during swing and a completely different walking pattern. So by constraining important DOFs, learned muscle activity patterns in the device might not result in a suitable overground walking pattern, which will decrease the likelihood of transfer of the relearned abilities to overground walking.

Moreover, the therapeutic spectrum reduces when DOFs that are characteristic of (impaired) human gait are constrained. Commercial devices actuate DOFs in the sagittal plane and focus on weight bearing and making an appropriate forward step. Training of balance control is not possible as the devices impose stability by constraining pelvic movements and hip abduction/adduction. Kollen and colleagues [16] demonstrated that improvement of balance control is the most important determinant in regaining walking ability, even more important than an increase in leg strength or decrease of synergies. So including the DOFs that allow the subject to actively practice his balance control during walking makes training in a robotic device more task specific and probably has a favorable effect on the outcome of robot-aided gait training.

21.2.2 Recovery as Well as Compensation Contributes to Functional Improvement

In clinical practice, a physical therapist focuses the therapy on achieving recovery of the paretic leg or on learning compensatory strategies that overcome the limitations due to impairments in the paretic leg. Recovery can be defined as restoring the ability to perform a movement in the same manner as it was performed before injury, whereas compensation can be defined as the appearance of new motor patterns resulting from the adaptation of remaining motor elements or substitution [17]. For example, in achieving an appropriate foot clearance during swing, a decreased ability to flex the knee can be compensated for by using a hip circumduction strategy constituting of increased hip abduction and pelvic rotation. However, most robotic gait training devices limit the therapeutic spectrum, since these devices focus on recovery to gain improvements in walking ability and do not allow to train compensatory strategies The robotic devices focusing on recovery direct their support at restoring a “normal” walking pattern and furthermore do not have the appropriate DOFs to allow or train compensatory strategies.

Currently, there seems no solid scientific evidence to favor the one recovery mechanism over the other. Several recent studies have demonstrated the importance of compensation in (the improvement of) functional walking ability in stroke survivors: stroke survivors using compensatory strategies can attain similar gait speeds as stroke survivors with “normal” movement patterns [18], a limited amount of generated propulsion (coordinated output) by the paretic leg does not necessarily restrict the gait speed [19], and improvements in walking ability during recovery are not accompanied by a restoration of the paretic muscle coordination patterns [20]. An often-heard argument against the use of compensation is that, in the long run, it might impede gains in other functional tasks. In the above-mentioned example, a circumduction strategy would, in all likelihood, not positively contribute to improving stair walking, whereas a recovery of knee flexion could. There is also accumulating evidence that targeted intervention results in recovery of the paretic leg: an intervention aimed at increasing ankle function results in specific increases of ankle power and an accompanying increase in gait speed [21]. So, recovery and compensation can both contribute to functional gains observed in stroke survivors. The contribution of each mechanism in bringing about functional improvements will probably depend on the patient’s impairments, their severity, and the time post-stroke.

To improve the outcome of robot-aided gait training, the devices should be directed not only at recovery but also at allowing and potentially even training compensatory strategies. This requires that the number of assisted and free DOFs of the robotic device should be larger than the number of DOFs of the task at hand, so the device provides redundancy. Attaining enough foot clearance while making a forward step can be regarded as a task with two DOFs. Allowing and/or actuating hip flexion and knee flexion suffices to perform the task. Yet adding hip abduction results in a redundant number of DOFs and makes compensatory strategies possible.

The need to allow compensatory strategies also has consequences for the control of robotic gait trainers. The control of the robot should allow the patient with sufficient freedom in how to move. This implies that we cannot define subject-independent reference trajectories for each DOF. Instead, these reference trajectories should be subject-dependent or should be defined in a coordinate system that allows the subject to choose his own strategy.

21.2.3 Active Training Required to Induce Cortical Plasticity

In the first instance, robotic gait training devices were developed for spinal cord–injured subjects and were designed to provide the spinal cord with the appropriate sensory information by imposing a normal walking pattern upon the subject. The legs were moved according to this pattern whether the patient was active or passive, and consequently, patients were not encouraged to actively participate. This approach was built upon scientific evidence from animal models that locomotor activity can be evoked by appropriately timed sensory information [22]. This information would drive central pattern generators, which are an ensemble of spinal cord neural networks that can generate basic rhythmical motor patterns involved in walking. Although similar central pattern generators likely exist in humans, there is growing evidence that the bipedal nature of human walking requires an important contribution of supraspinal structures in controlling walking. This evidence could be gathered through advances in brain imaging and electrophysiological techniques that allowed investigation of supraspinal control of walking. Miyai and colleagues [23] measured the brain activity of healthy subjects during gait and showed that the medial sensorimotor cortices and the supplementary motor cortical areas were involved in the control of walking.

The supraspinal involvement in the control of walking implies that brain plasticity can contribute to improvements of walking ability, which has major consequences for the design of (robot-aided) gait training. Indeed, several studies using different technologies showed that changes at a cortical level and also on subcortical level correlated with locomotor recovery in stroke survivors [24–26]. Also, in spinal cord injury, subject brain plasticity contributes to locomotor recovery. After 3–5 months of treadmill training, SCI subjects showed an increase in evoked muscle responses from TMS to the leg area of the motor cortex that were related to locomotor recovery and could not be explained by increased spinal excitability [27].

The process underlying this brain plasticity/reorganization is driven by self-generated activity, which stresses the need of a subject to actively participate in the training and not being passive. The importance of self-generated activity over passive guidance was emphasized in a study by Lotze and colleagues [28] in healthy subjects. They showed that a training period consisting of voluntary induced (active) wrist movements resulted in larger performance improvement and cortical reorganization than passively induced movements. These results were later replicated for the lower extremities by Perez and colleagues [29], who also showed that not just repetitively performing a movement induces cortical plasticity but that the generated activity should be part of a skill. They compared the changes in corticomotor excitability in subjects who received skill training consisting of a pursuit tracking task by performing ankle plantarflexion and dorsiflexion, passive training in which subjects were assisted in the pursuit tracking task, or nonskill training consisting of just voluntary performing plantarflexion and dorsiflexion. Only subjects receiving the skill training showed an increase in cortical excitability that was accompanied by an improved performance.

These studies show that neurological patients should be encouraged to actively contribute in robot-aided gait training (and not to rely on the robot) in order to facilitate plasticity-induced improvements in walking ability. The tasks given during training should be clearly related to the skills that are important in walking, like balancing and foot placement. Additionally, the patients should not only be promoted to actively participate, but they should also be allowed to experience errors in the task execution as in the end task execution errors drive motor learning [30].

21.3 Mechanical Design of LOPES

Robotic gait training devices differ widely in their (actuated) DOFs and how they are controlled (see Table 21.1). The choice of which DOFs to restrain, actuate, or let free depends on the underlying view on neurorehabilitation and on the nature and control of human walking. Arguments can be given for more DOFs, but these are balanced by the consequence that the device will be more complex and expensive. At this moment, there is no solid evidence which DOFs to actuate or not since no comparative studies have been performed between devices with different DOFs. In the next paragraphs, we will provide the arguments for the chosen DOFs of LOPES.

The DOFs of LOPES and how they are actuated (see Table 21.1 and Fig. 21.1) are chosen in such a way that they allow unhindered walking in the device (transparent mode), allow the use of compensatory strategies and to selectively support the essential aspects of walking. A prerequisite for selective support is that the device itself is transparent. The transparent mode is needed at the end of the training program, when the subject only requires little support, since the device should resemble normal walking as close as possible to facilitate the transfer of the learned abilities to overground walking. Another argument for the importance of this transparent mode is that in hemiparetic gait, only the affected leg needs support while the unaffected leg should be able to move freely. We will first exemplify the

Fig. 21.1 Subject attached in the first prototype of the LOPES device. The eight actuated DOFs are schematically indicated

choice of the DOFs in the light of the requirement that the essential aspects of gait should be selectively and partially supported.

When determining the essential aspects that need to be supported, we paid great attention to the inherently unstable dynamics of walking. Walking can be considered as controlled falling in a desired direction. The lateral and forward foot placement is used to stabilize gait and control balance [31, 32]. Therefore, hip flexion/extension and hip abduction/adduction are actuated. Also, horizontal pelvis motions are actuated as constraining or reducing pelvis motion would externally stabilize gait. Different studies have shown that constraining pelvis movements affects foot placement and increases trunk motion [33, 34]. Other essential aspects that need to be supported are foot clearance during swing and weight bearing during stance, which require actuation of knee flexion/extension. Also, the propulsion is an important aspect of gait. Hip extension during initial stance contributes to propulsion [35], but the main contributor is plantar flexion at the ankle. Still, we decided not to actuate the ankle to reduce mass and complexity of the device. Different actuated orthoses have been developed to specifically support the ankle during gait [36, 37]. Future clinical testing with these devices has to show the additional value of incorporating ankle plantar flexion.

The DOFs needed to support the essential subtask also suffice in meeting the other abovementioned requirements. The inclusion of the hip abduction/adduction degree of freedom allows for using one of the most often used compensatory strategies, the hip circumduction. The total set of DOFs allows all major movements of gait to be made with the device, so walking with the device can resemble walking outside the device as long as the dynamics of the exoskeleton does not influence walking with the device too much.

Another important requirement for the mechanical design of LOPES is related to the dynamics of the exoskeleton. For LOPES, and generally for force-controlled devices, it is important to minimize the inertia of the device since control algorithms can only partly compensate for the inertia. Therefore, we build a lightweight exoskeleton that has the heavy motors and gearing detached from the exoskeleton. Newly designed Bowden-cable-driven series elastic actuators are used to transmit the mechanical power of the motors via Bowden cables to the actuated joints [38]. This actuation also resulted in the required high torque control bandwidth that is needed for impedance-controlled devices. The torque control bandwidth of LOPES is 16 Hz [39].

21.4 Control of LOPES

The control of robotic devices greatly determines whether patients are encouraged to actively participate in the training but also whether patients are allowed to use alternative movement strategies. The first generation of robotic gait training devices mainly used position control to move the patient's legs through a prescribed gait pattern, irrespective of the patient's self-generated activity, and not allowing the patient to use compensatory strategies. To increase the active participation, more and more robotic devices control the interaction forces by using impedance or admittance control algorithms. Mostly, reference position trajectories are still used in this approach to determine the amount of force to apply. The control of interaction forces brings along new challenges, as how and when to support the patient, and to decide how large the amount of support should be.

By controlling the interaction forces, the amount of support can be adapted to the patient's needs and abilities: the robot can still be very stiff and practically enforce a gait pattern when the patient is not capable of generating any appropriate activity, can be very compliant and move with the patient when the patient is generating the appropriate movement, and everything in between. One of the biggest challenges is how to determine the appropriate amount of support for each specific patient. Different algorithms have been developed to automate this process. Emken and colleagues [40, 41] developed and evaluated an error-based algorithm with a forgetting factor based on motor adaptation experiments in healthy subjects. One term in this assist-as-needed algorithm increases the support when deviations from the reference trajectories become larger, whereas a second term gradually reduces the support from step to step. The resulting support is the equilibrium between these two terms. They showed that the support was shaped to the patient's specific needs. An appropriate choice of the parameters of this algorithm would not only assure automatic adaptation of the support but would also prevent reliance on the robotic support to occur. Hitherto, this latter aspect has only been shown in experiments with healthy subjects and in simulation studies and not in experiments with neurological patient.

Another challenge is in the timing of the robotic support. When using reference trajectories, these trajectories should be synchronized with the movements of the subjects. Lowering the stiffness/impedance increases the likelihood that the reference and actual movement are not in phase. This phase difference can grow rapidly over different steps and turns the robot's supportive forces into uncomfortable and unwanted perturbations. Different algorithms have been proposed and evaluated to synchronize the robot's actions with the actual movements. Aoyagi and colleagues [4] proposed and demonstrated the appropriate working of an algorithm that continuously adapts the "replay" speed of the reference trajectory to minimize the difference between the timing of reference pattern and the patient's movements. Duschau-Wicke and colleagues [42] proposed a method in which variation in timing is allowed within a specified time window. When the timing error exceeds the window, the robot will apply additional torques to slow down or speed up the movements of the patient.

The control approach is also important in allowing or even training alternative movement strategies, given that the used robotic device provides redundancy in the DOFs. Most robotic devices are controlled at a joint level, and reference patterns are also defined at a joint level. This complicates the definition of reference patterns for alternative movement strategies. Although compensatory strategies can be classified into a limited number of widely used strategies, there still is considerable variation between patients within a "class," as the actual strategy is highly dependent on the patient's impairments. As such, it is hard to define appropriate reference patterns that can be generally used but also to define subject-specific patterns. Still, the latter can be done by using a teach-and-replay approach [4]. In this approach, the robot is first controlled in such a way that it does not actively assist the movement. The necessary guidance is provided by a physical therapist who moves the leg through the desired pattern, and the robot records these movements. Subsequently, this recorded trajectory is used as a reference to what amounts to an endless repetition of the therapist's actions.

For LOPES, we developed and applied an alternative approach to tackle the previously described challenges. The core of this approach is that we divide human gait in different subtasks, and the performance on each of these subtasks is evaluated and controlled separately. These subtasks are: attaining sufficient foot clearance during swing, making a forward step, weight bearing, weight shifting, stance preparation, and balance control. This approach is called selective subtask control. Each subtask is controlled in parallel by using virtual models, like virtual springs and dampers, which are defined between the actual performance and the defined reference on the concerned subtask. The forces in these virtual models are transformed into the required robotic joint torques which are exerted by LOPES on the human limb. Recent simulation and experimental studies [43, 44] have provided evidence that humans also control walking in a modular approach as the muscle activity during walking can be decomposed in different modules associated with different subtasks.

In our approach, the amount of support can be adapted to the patient's needs in two different steps (see Fig. 21.2). First, the therapist selects the subtasks, which are impaired in the subject, to be controlled by LOPES [45]. Second, the amount of support in each of the controlled subtasks is adapted to the patient's needs by using an adaptive algorithm. In this way, patients are supported as much as necessary on the impaired aspects of gait while they have to generate all the activity for the unimpaired aspects by themselves [46]. Synchronization problems are prevented because the support is gait phase dependent. This means that a specific subtask is only controlled during the phases in which the subtask should be performed (see Fig. 21.2), and the control is actually reset for every gait cycle. The control on a subtask level also leaves room for compensatory strategies. Subjects can use different strategies to accomplish a certain subtask as the reference pattern is not defined on a joint level but on a subtask level. For instance, the patient can use a hip circumduction strategy instead of regular knee flexion to get enough foot clearance. If by using this strategy the patient indeed succeeds in attaining appropriate foot clearance, no support will be provided. If not, the support can either be directed at improving knee flexion or at using a compensatory strategy.

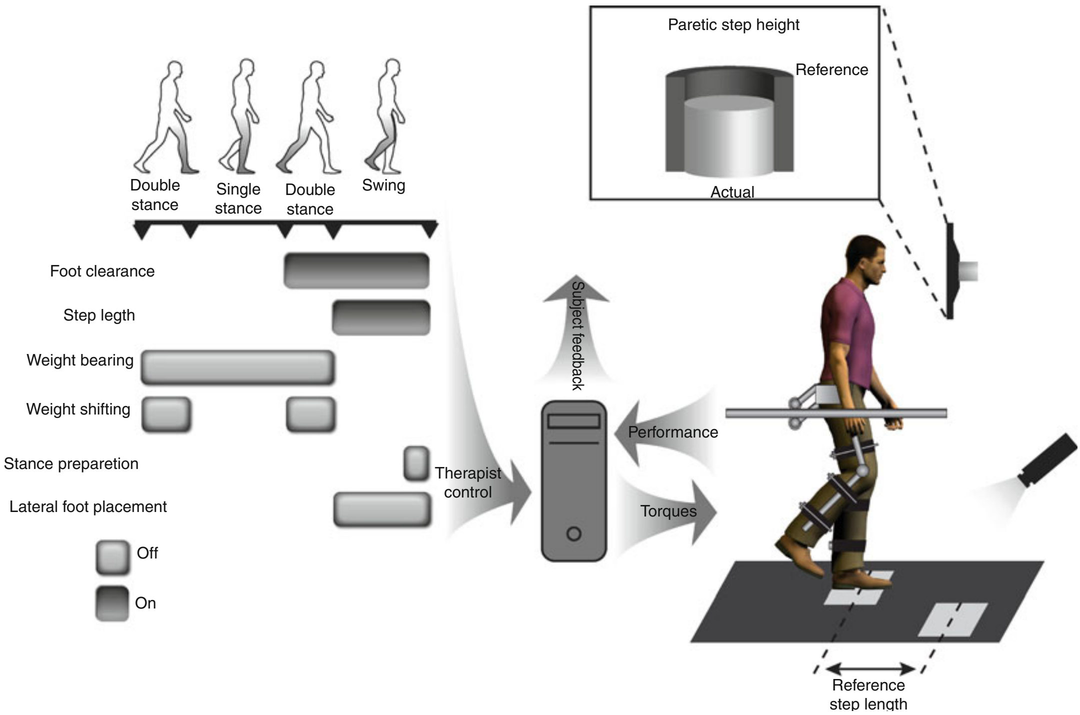

Fig. 21.2 Schematic overview of the used approach to selectively support different subtasks of gait with LOPES. This control allows for an intuitive control for the patient and therapist. The therapist decides on which aspects of gait the patient needs support. Based on this selection, the implemented control algorithms calculate the required supportive torques. The level of support is automatically adapted to minimize the robotic support and maximize the patient's participation by using an assist-as-needed algorithm. The reference or target values for each subtask are displayed on a screen in front of the patient or on the treadmill (by using a beamer). To stimulate the active participation of the patient its actual performance on each subtask can also be displayed

Another advantage of using selective control of subtasks is that it allows to provide intuitive feedback about the performance on each of the subtasks to the subject and therapist and that target values on each of the subtasks can be presented to the subject (see Fig. 21.2). Our experience is that setting the targets and providing feedback on gait parameters like step length and height are easier to interpret for patients as well as therapists than feedback in terms of joint angles or torques.

21.5 Experience with and Feasibility of LOPES

Only providing assistance as the patient needs it, not only requires that the robot is able to provide the necessary assistance but also that the robot does not hinder the motion of the subject when no assistance is required. As a first step in implementing LOPES into gait training, we evaluated this latter requirement by comparing the gait parameters, kinematics, and muscle activity of ten healthy subjects while walking with LOPES attached to their pelvis and limbs and while walking freely on a treadmill [47]. In this study, LOPES was controlled to provide no assistance (transparent mode). Overall, the patterns of the joint and segment movements and those of muscle activity while walking with LOPES resembled those of free walking. However, various changes did occur, which could be mainly ascribed to the mere fact that the attached exoskeleton added inertia to the subject's legs which needed to be accelerated and decelerated by the subject. Muscles involved in accelerating the leg during initial swing, like the rectus femoris, and muscles involved in decelerating the leg during terminal swing, like the biceps femoris, both showed an increase in activity (see Fig. 21.3). In addition, the added inertia resulted in a decreased knee flexion during swing which on its turn likely induced the increase in tibialis anterior activity to achieve appropriate foot clearance. Apart from the inertia of the exoskeleton legs, the subject experienced some resistance in moving the pelvis, which caused a significant increase in the frontal trunk rotations. All in all, the results were satisfactory in that the walking pattern with the device was similar to the normal walking pattern. However, they do show the importance of reducing the inertia of the exoskeleton or developing algorithms to compensate for it when one wants to achieve unhindered walking in a robotic device.

In a subsequent study, we determined whether ambulatory chronic stroke survivors were able to make use of the DOFs of the device. The included stroke survivors had a decreased amount of knee flexion during the swing phase, which is an often-reported gait abnormality in stroke survivors and is also referred to as stiff knee gait. They walked with LOPES when again it was controlled to provide no assistance, so they were not forced to a certain pattern and were free to adopt their own walking pattern. When walking in LOPES, subjects indeed showed a marked lower knee flexion range in the paretic leg compared to the nonparetic leg (see Fig. 21.4). Most subjects compensated for this by using a hip circumduction strategy which was reflected in the large amount of hip abduction during swing. There seemed to be a trend in that the lower the knee flexion range, the larger is the amount of hip abduction. Subjects using a hip circumduction strategy in LOPES also used this strategy while walking overground. These results demonstrate that subjects can use their own movement strategy in the device and that they experience the result of their self-generated activity.

The feasibility of the selective support of subtasks has been demonstrated in experiments with healthy subjects for several subtasks, among which are attaining sufficient foot clearance, making a (larger) step, and weight bearing. In these experiments, subjects walked with LOPES, and the support on a specific subtask or combination of subtasks was switched on during selected steps, whereas during the other steps and on the other subtasks, no support was provided. In general, the feasibility was assessed by determining how well the set reference values were attained and how the support affected the remaining of the walking pattern. For the step height and step length, the reference values were set at a 15% increase with respect to their normal values. The support of step height resulted in an increase of the step height that was caused by an increase of the knee flexion during swing (see Fig. 21.5). The use of a stiff virtual spring in the controller resulted in a significant closer approach of the reference value compared to using a compliant spring. This support was selective in that it did not affect the other basic gait parameters like step length or cycle time. The support of step length resulted in a less selective effect as not only the step length showed a significant increase but also the step height showed a significant decrease. The accompanying decrease in step height could be explained from the exerted robotic torques to increase the step length, as to increase the step length, the robot exerted hip and knee extension torques. The support of step length also showed to be less dependent on the used virtual stiffness. When the support of step length was combined with the support of step height, the effects of the separate support algorithms were combined, and the increase in step length was accompanied by an increase in step height.

Weight bearing during stance can also be considered as a subtask of walking. Using a robotic gait trainer to support weight bearing might have considerable advantages over typical overhead suspension systems. These latter systems are often used in gait training to provide the patients with the required amount of body weight support, but do have some disadvantages. Over the last years, different studies [48, 49] have demonstrated that this form of body weight support considerably influences the spatial, temporal, and kinematic gait parameters in healthy subjects. Although some more advanced systems [50] allow the modulation of the amount of support

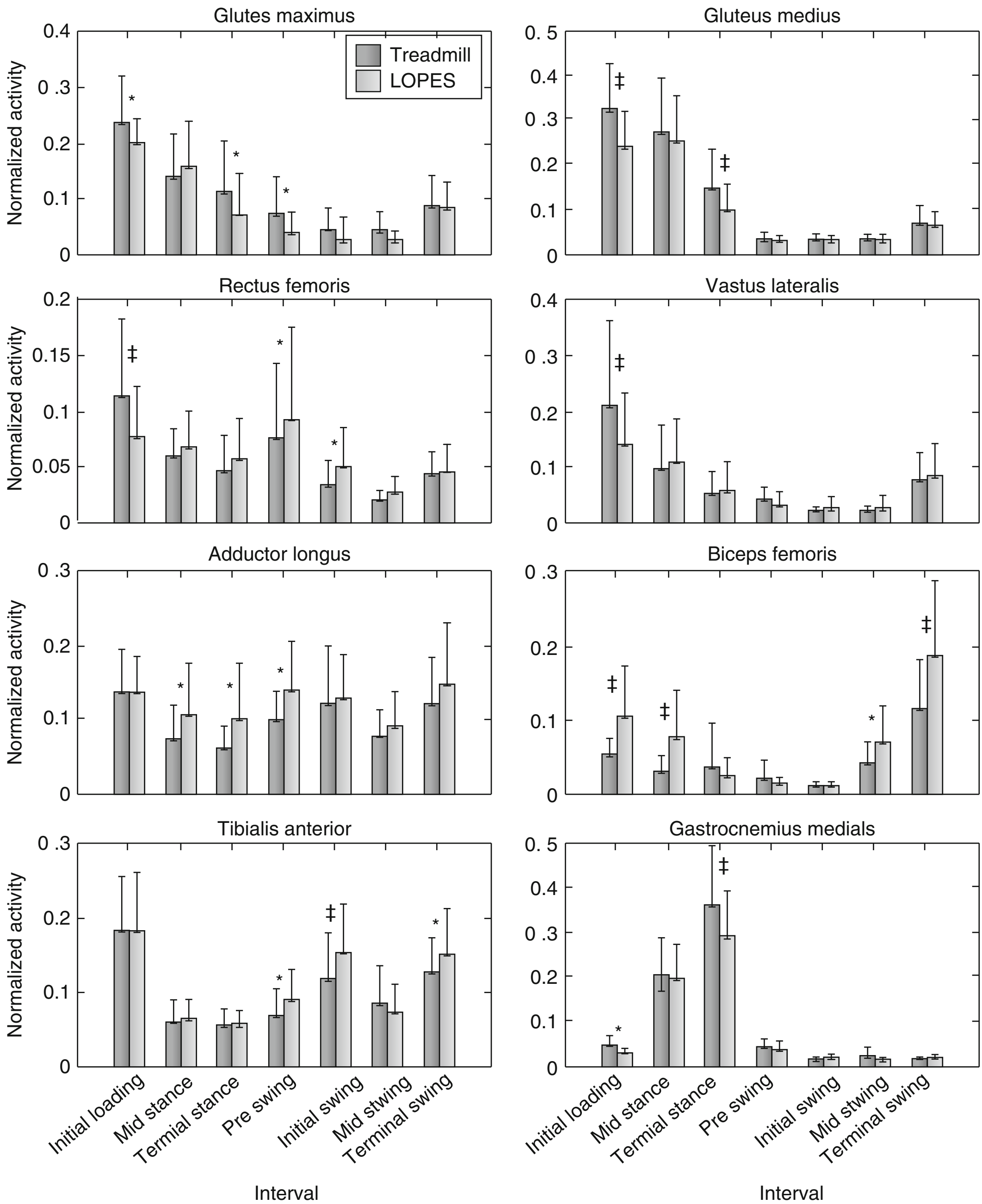

Fig. 21.3 Muscle activity of healthy subjects walking in LOPES when it is controlled to provide no assistance. Mean normalized integrated activity for eight leg muscles over seven gait intervals for LOPES walking and treadmill walking. The *vertical bars* indicate the standard deviation over the different subjects. Significant differences between LOPES walking and treadmill walking are indicated with an * for $p<0.05$ and with a ‡ for $p<0.001$ (Reprinted from van Asseldonk et al. [47]; with permission. © 2008 IEEE)

between the different legs, most systems support an equal amount of body weight support during stance of both legs, whereas hemiplegic subjects only need the support during the stance phase of the affected leg. Additionally, typical systems do not provide a force in the pure vertical direction

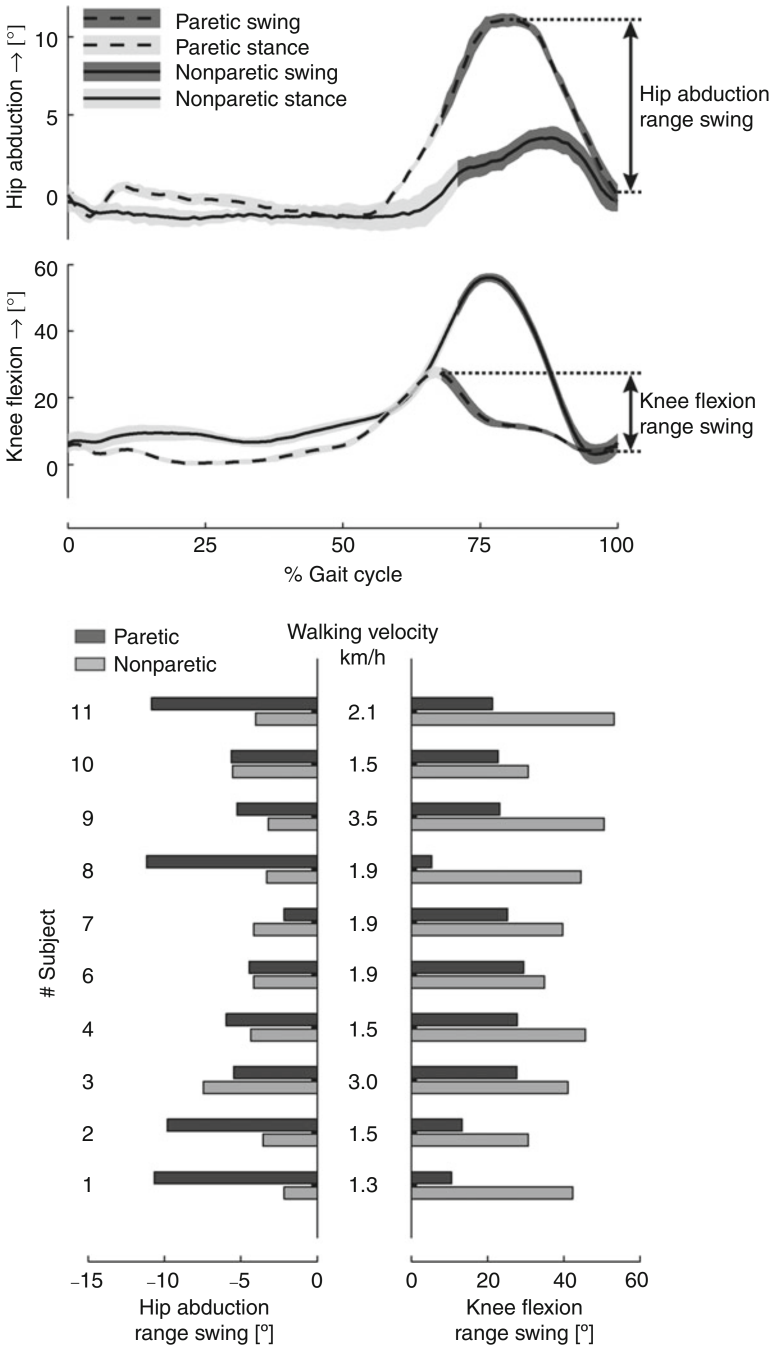

Fig. 21.4 Compensatory strategies of chronic stroke survivors walking with LOPES. The *upper graphs* show averaged trajectories of hip abduction/adduction and knee flexion/extension of a typical chronic stroke survivor (subject 11) walking with LOPES, which is controlled to provide no assistance. The *shaded areas* indicate the standard deviation. The *lower graphs* show averaged ranges of hip abduction and knee flexion during the swing phase of ten ambulatory, chronic stroke survivors with stiff knee gait walking in LOPES

but also in the horizontal plane that helps subjects to maintain their balance. This implies that the amount of support on weight bearing and balance control cannot be independently varied, whereas the amount of impairment on each of these tasks varies widely within and between subjects.

The aforementioned disadvantages can be overcome by using a robotic exoskeleton. We have assessed the feasibility of a control algorithm to support the subject in weight bearing by exerting torques on the joints to overcome the gravitational torques and to prevent knee buckling [51]. This

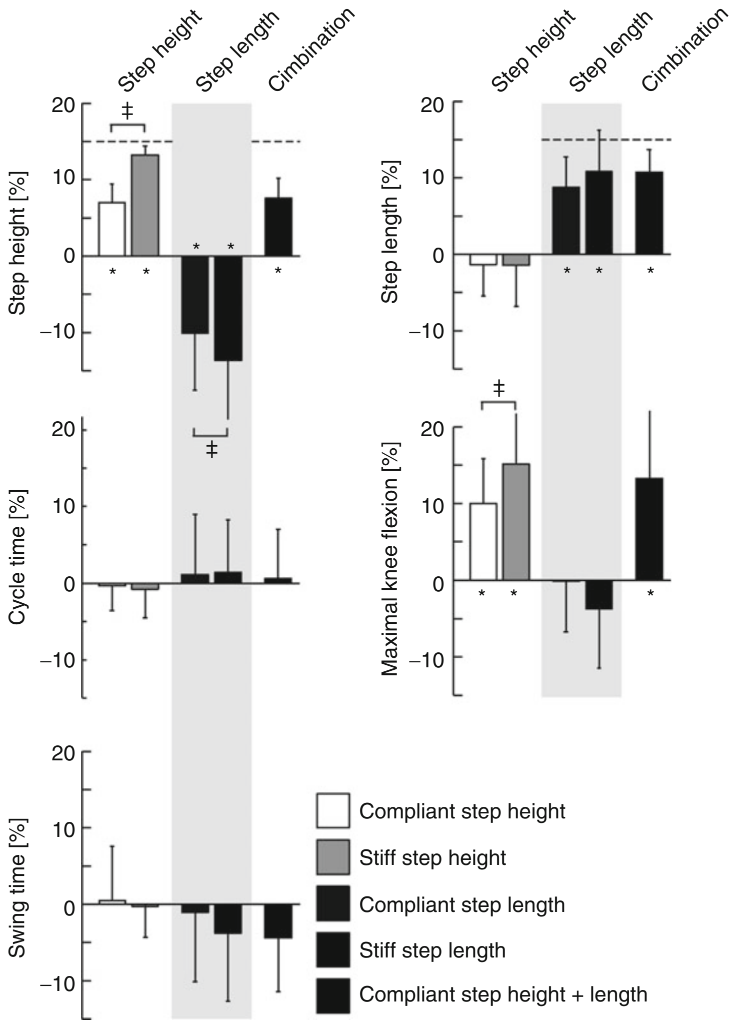

Fig. 21.5 Effects of exposure to selective subtask control on different spatiotemporal gait parameters. The *bars* indicate relative average (across six subjects) changes in gait parameters with respect to a baseline measurement. The *vertical lines* indicate the standard deviation. Subjects were being exposed to selective control of step height with a compliant (600 N/m) and stiff (1,200 N/m) virtual spring, selective control of step length with a compliant (400 N/m) and stiff (800 N/m) virtual spring, and a combination of the step height and step length support with compliant springs. The reference values during support were set to 115% of the baseline values. An * indicates significant difference with zero (the value is changed due to the support), and a ‡ indicates a significant difference between the compliant and stiff condition. The *dashed gray horizontal lines* indicate the set reference values

algorithm allows for independent control of weight support during stance of the different legs and does not interfere with balance control. Results showed that the algorithm was effectively supporting weight during loading as the muscle activity of important knee extensors decreased, whereas the pattern and range of angular movements resembled those of walking without the support.

All in all, these results showed that the different aspects of gait can be supported separately but not always selectively. A combination of selective controllers can be used to provide support on multiple aspects or to provide support on one aspect and set a boundary condition on another aspect. By selecting subtasks which require support, the robotic assistance can be adapted to the capabilities of a subject. However, also within a subtask, the amount of support needs to be adapted to fit the needs of the patient. The support should be such that large errors are prevented and safe walking is guaranteed and such that small errors and variation over steps are allowed.

To adapt the support within a subtask, we incorporated the error-driven adaptation algorithm of Emken and colleagues [41] in the selective control of step height [46]. The resulting algorithm modified the virtual spring stiffness at each percentage of the gait cycle based on the experienced error in the previous steps. We evaluated this algorithm in ambulatory chronic stroke survivors. These stroke survivors did not need the

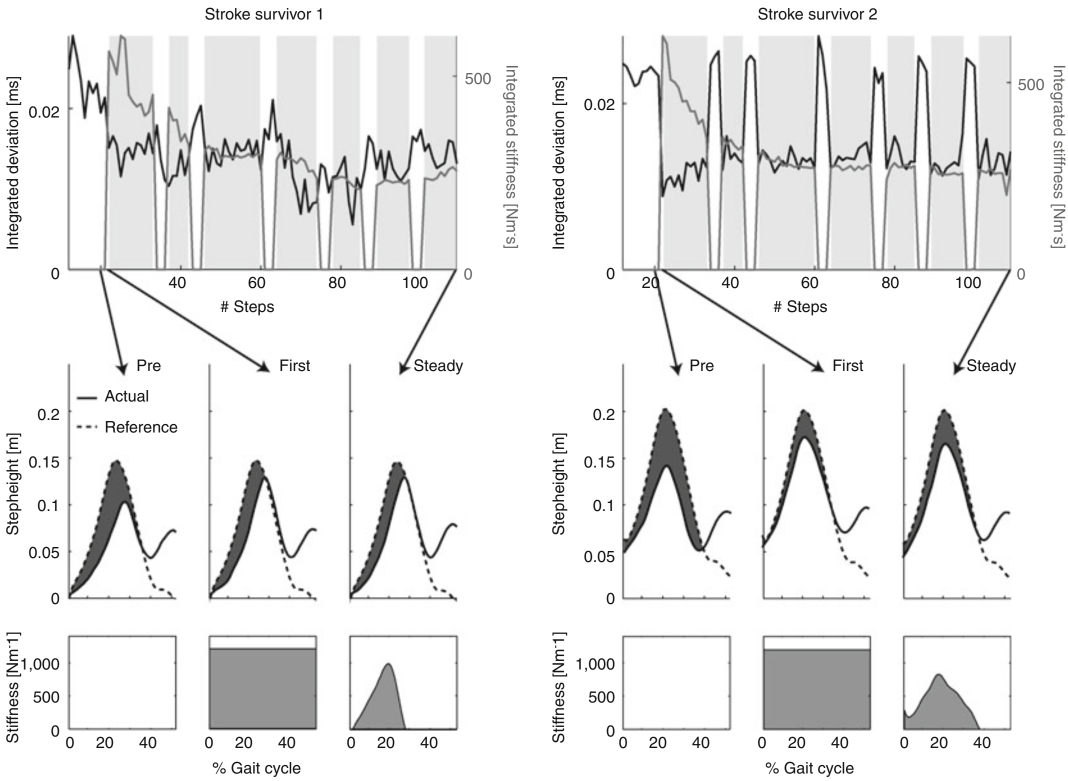

Fig. 21.6 Shaping of the virtual stiffness of the step height controller in two ambulatory chronic stroke survivors. The *left* and *right* set of graphs shows the responses for two different chronic stroke survivors. The *upper row* shows the course of the deviation from the reference (*light gray line* and *axis*) and the stiffness (*dark gray line* and *axis*) over multiple steps in a walking trial. The support is turned on after 20 steps and turned off for three steps after random intervals. The *shaded vertical bars* indicate the periods in which the support was turned on. The measures for the error and stiffness are obtained by integrating the *shaded area* indicated in the middle and lower row of graphs over time for each separate step. These graphs show the actual and reference ankle height (*middle row*) and virtual stiffness (*lower row*) as a function of the gait cycle for the step preceding the first exposure (stiffness is zero), for the first step of exposure (stiffness is constant, no shaping), and for a step when subjects walked for 70 steps with the support

robotic support to walk; the provided support was purely aimed at increasing their foot clearance. The results showed that the combined algorithm was effective in adapting the amount of support to each subject's capabilities (see Fig. 21.6). The profile of the virtual spring stiffness (stiffness versus percentage of the gait cycle) and the exerted robotic support were shaped to the initial deviation of the actual ankle trajectory from the reference trajectory.

Interestingly, subjects responded quite differently to the provided support, which stood out clearly by making use of "catch steps." In these steps, the subjects were not receiving any support, and these trials were randomly interspersed among the steps with support. Some subjects (see subject on the right in Fig. 21.6) did not take over the robotic support by improving their walking pattern. In these subjects, during the catch trials, the deviation of the step height from the reference increased to presupport values. Still, the subjects did not rely on the support, since the deviation did not increase above the presupport values. Other subjects utilized the robotic support (see subject on the left in Fig. 21.6) to improve their own performance. In these subjects, the integrated error during the catch trials decreased in comparison to the presupport errors (see for instance catch trial around step 73). In short, the adaptive algorithm automatically adjusts the amount of support to the capabilities and the actual performance of the subject for the specific subtask; this reduces the need for the therapist to set the amount of the support on a trial and error basis. However, currently, the used parameters in the adaptive algorithm are not set specific to the subject, which would also decrease the chances of reliance on the support. The identification of the appropriate parameters is very cumbersome in neurological patients, and new methods need to be developed to make this identification possible.

The next step in the development of LOPES was to perform a first explorative training study in a small group of ambulatory chronic neurological patients. Five chronic stroke survivors whose gaits were characterized as stiff knee gait participated in a 6-week training program. During the training, the subjects received support using the previously described adaptive support of step height. The provided support was directed at facilitating recovery of function in the paretic leg. All subjects showed a marked increase in walking velocity during training. Yet, there was only limited transfer of this gain to overground walking (see Fig. 21.7). A larger gain in speed during training compared to overground walking has also been reported for body weight support training [35]. Still, the limited transfer might also indicate that walking in LOPES does not yet resemble overground walking enough. During training, subjects were stabilized as they were holding the side bars, and the dynamics of the device provides some stabilization, whereas during overground walking, this kind of stabilization is not provided. In two of the five subjects, the training resulted in a considerable increase in knee flexion during swing (5° or larger) in overground walking. Whether a subject showed an improvement in knee flexion or not was not clearly related to the walking ability at the start of the training or clinical measures of motor functioning like the leg portion of the Fugl-Meyer. The small number of patients included and the variation in effect between subjects do not allow drawing firm conclusion about the added value of the selective robotic support on promoting recovery of function. Still, as changes in overground walking velocity were rather small, and only two subjects showed an increase in knee flexion, we could argue that it might be more efficient in some chronic stroke survivors to direct the provided support on the use of compensatory strategies instead of on recovery of knee function to improve walking velocity.

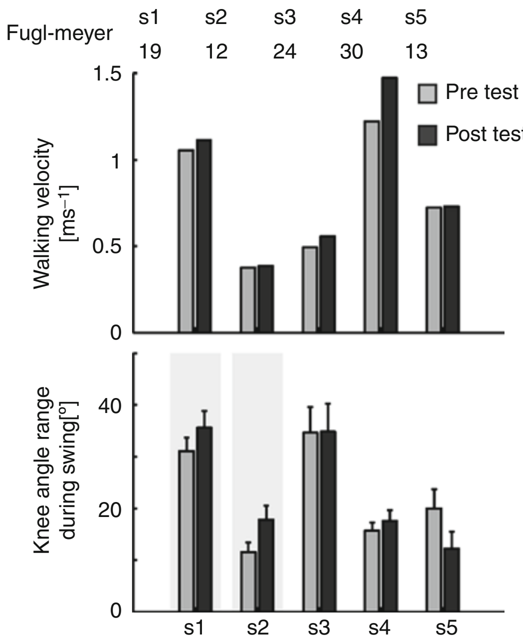

Fig. 21.7 Effect of training with selective support of step height on overground walking velocity and knee angular movement in chronic ambulatory stroke survivors. *Vertical bars* indicate the standard deviation. Subject s5 experienced a serious fall in a home situation during the training period but was able to complete the training

21.6 Current Developments and Ongoing Testing

From the results we obtained so far with LOPES, it can be concluded that the walking pattern while walking with LOPES in the transparent mode resembled overground walking, that patients utilize the redundant DOFs to make use of compensatory strategies, that the support on the level of subtasks is feasible, and that the amount of support can automatically be adapted to the specific needs of the patients.

These results are encouraging; however, LOPES is still under development, and different aspects need further improvement, and new features need to be developed. First, the mechanical design and control of LOPES should be improved to provide less unwanted stabilization. The external stabilization provided by LOPES can largely explain the observed differences between overground walking and walking in LOPES and the limited transfer of the improvement in speed during training to overground walking as observed in the clinical trial. Second, we will extend, refine, and test the controllers to provide selective subtask control. We will pay special attention to controllers that provide support in balance. Third, we are developing feedforward controllers. Currently, the provided support is realized with feedback controllers, but these do not suffice for severely affected patients. Fourth, the observed difference in responses between patients to the currently implemented adaptive algorithm suggests that further optimization and individualization of these adaptive algorithms and their parameters is needed.

Effect studies in (sub)acute patients have to prove that selective support of intuitive subtasks according to the minimal robotic intervention principle indeed increases the active participation of patients and results in functional improvements that are at least as large as those obtained with conventional training. To perform these effect studies, LOPES is now being redesigned to make it suitable and available for rehabilitation clinics.

21.7 Perspectives

The application of robots in gait training is a relatively new development. Randomized clinical trials showed that conventional therapy outperforms the first generation of robotic devices. Recent insights and developments resulted in new devices and modifications of existing devices that overcome some of the limitations of the first generation of robotic gait trainers. In designing and controlling robotic devices, choices have to be made. We made these choices to improve on the task specificity, active participation, and facilitating different recovery process, whereas other researchers and companies might want to improve the training on other aspects. Clinical trials need to prove that the next generation of robotic gait training devices results in larger functional improvements and/or faster improvements. Comparison of the outcome of the clinical trials with the different devices should provide us with insight in which training aspects are the key elements in facilitating functional improvement.

In the end, robot-aided training should be tailored to each patient's specific impairments, capacities, and prognosis. This requires objective and quantitative measures of the impairments and capacities. The unique features of robotic gait training devices can be used to obtain (some of) the measures. So, robotic gait training devices can be used not only to apply the training but also to predict whether the training will be effective and what the content of the training should be.

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