Restitution and compensation in the recovery of function in the lower extremities of stroke survivors

design of evaluation and training methods

Edwin van Asseldonk

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RESTITUTION AND COMPENSATION IN THE RECOVERY OF FUNCTION IN THE LOWER EXTREMITIES OF STROKE SURVIVORS

DESIGN OF EVALUATION AND TRAINING METHODS

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contents

chapter 1

general introduction

1.1 background

Stroke is a leading cause of long-term disability in the Netherlands and the rest of the industrialized countries. Either by obstruction of a blood vessel supplying blood to (a part of) the brain, or by rupture of one of those blood vessels, brain cells get deprived from oxygen, which results in irreversible damage to these specific brain regions. When brain areas that control movement are involved, this results in impaired execution of the movements on the side of the body opposite (contralateral) of the side in which the damage to the brain occurred. Muscle weakness (paresis), which can be considered to be a direct consequence of the decreased volitional drive from the damaged brain, is considered to be one of the major factors compromising motor function in the upper (Wagner *et al.*, 2006) and lower extremity (Nadeau *et al.*, 1999; Kim & Eng, 2003). In addition, other impairments like changes in reflex excitability (Sinkjaer & Magnussen, 1994; Mirbagheri *et al.*, 2007) or the inability to selectively control movements at the separate joints (Beer *et al.*, 1999; Dewald & Beer, 2001) can hinder the execution of movements in different degrees (Beer *et al.*, 2007; Dietz & Sinkjaer, 2007; Sukal *et al.*, 2007).

During the rehabilitation of stroke patients, restoration of motor function in reaching, gait and balance are important goals. Presently, more than nine neurological treatment approaches exist that are used in post-stroke treatment (Van Peppen *et al.*, 2004b). All these interventions differ considerably in their approach to achieve functional gains. For instance, Bobath (or NDT) which is the most often used approach in the Netherlands (Van Peppen *et al.*, 2007) concentrates on inhibiting abnormal tone of the muscle and synergistic-dependent motor control to restore the pre-stroke movement strategies in performing functional tasks. Other approaches place less emphasize on restoring normal movement strategies and even facilitate other movement strategies if this could lead to better functional improvements. Despite these differences in facilitating recovery, the conventional therapy interventions have resulted in similar functional outcomes (Van Peppen *et al.*, 2004a). Although the functional end levels are the same, there could be a difference in how these end levels are reached. Remarkably it is unclear if the functional improvements are obtained in different ways, that is, through the use of normal movement strategies or through the use of alternative movement strategies (Lennon *et al.*, 2006). Generally, there is a lack of insight in the underlying recovery mechanisms in the field of neurorehabilitation (Kwakkel *et al.*, 2004a).

The specific recovery mechanisms are largely unknown, as appropriate methods for reliably quantifying these mechanisms have long been or are still unavailable. The development of these methods is a requisite for increasing the understanding. This thesis will deal with the development of these methods. In the following paragraphs, an overview will be given of the already available methods, of the difficulties that are encountered in developing methods for quantifying the recovery mechanisms in the lower extremities and of the reported evidence for the different recovery mechanisms.

1.2 definition of restitution and compensation

Quantifying the recovery mechanisms that underlie functional improvement have only recently attracted considerable attention. It has been argued that two distinct mechanisms account for post-stroke improvement. Restitution (also called true recovery) leads to a return of pre-lesion movements and/or function of the affected limb. Compensation is the result of emergence of new movement strategies that differ from the original (Cirstea & Levin, 2000; Kwakkel *et al.*, 2004a). For example, if impairments in the control of the paretic arm joints hinder a stroke patient in reaching to an object, a rotation of the trunk can be used to compensate for the impairments in the arm (Cirstea & Levin, 2000; Michaelsen *et al.*, 2004). The aforementioned example exemplifies that compensatory strategies are dependent on the redundancy of the musculoskeletal system to combine individual joint movements in achieving a motor task.

1.3 quantifying restitution and compensation

The relative contribution of restitution and compensation in recovery is largely unknown. Distinguishing between restitution and compensation requires at least detailed analysis of movement. These mechanisms cannot be deduced from the functional scales or impairment scores that are mostly used to evaluate the effectiveness of a therapy/training. In the functional scales the ability of a subject to perform a specific task (walking, reaching and turning) is scored on an ordinal scale and they do not distinct between movements made using compensation strategies or movements made using restituted movement patterns. Furthermore, the recovery mechanisms cannot be directly inferred from measurements of impairments, such as muscle weakness in the paretic arm. Weakness of a specific muscle may provoke the use of compensatory strategies (Cirstea & Levin, 2000; Michaelsen *et al.*, 2004), however this does not imply that the use of a compensatory strategy can automatically be derived from a specific impairment.

The use of a compensation strategy can be inferred from a detailed single measurement, however measurements in a longitudinal design are necessary to deduce whether during recovery the compensatory strategy is replaced by the original movement pattern (restitution) or whether the patient holds on to the compensatory strategy. In the following paragraphs, the different methods to quantify compensation in reaching, walking and balance control will be indicated. Subsequently, in paragraph 1.4, the results of longitudinal application of the methods will be described.

1.3.1 upper extremities

The arms are mostly used for reaching and grasping. The goal during reaching is simply to bring the hand from a starting position to the goal position. From the movements of the hand, parameters can be extracted that quantify the motor control. Recently several studies have used this approach to quantify the disturbed motor control in reaching of stroke survivors (Rohrer *et al.*, 2002; Wagner *et al.*, 2006) and to assess how training/therapy affects the motor control (Rohrer *et al.*, 2004; Lang *et al.*, 2006; Colombo *et al.*, 2007; Dipietro *et al.*, 2007; Wu *et al.*, 2007). However, measurements of hand position alone do not suffice in determining whether compensation strategies are used or not. For this purpose, the kinematics of whole arm needs to be quantified. By comparing the movements of stroke survivors with those of healthy subjects, who make rather typical joint movement patterns, the use of different strategies can be derived.

Levin and colleagues extensively studied the use of alternative movement strategies in chronic stroke patients. They showed that when the movements from the elbow and shoulder were impaired, the subjects incorporated movements of the trunk to extend the reach of the arm. The amount of trunk movements was highly correlated with the motor function of the impaired subjects, that is mildly impaired subjects used healthy movement patterns, whereas the moderately to severely impaired subjects incorporated the use of trunk movements (Cirstea & Levin, 2000; Levin *et al.*, 2002). In a subsequent study (Michaelsen *et al.*, 2004), they showed that compensatory trunk movements were also used for orienting the hand in grasping. Apart from assessing the importance of compensatory movements, they also investigated the modifiability of the movement strategies. Interestingly, they (Michaelsen *et al.*, 2001) demonstrated that by restraining movements of the trunk, chronic stroke survivors were able to employ movement ranges at the elbow and shoulder joints, that they would not normally use. So, by restraining trunk movement, the impaired subjects showed movements patterns with a closer resemblance to normal patterns. These results suggest that using a trunk restraint during training could be an efficient way to aim for restitution of the original movement patterns during training.

The aforementioned studies show that compensatory strategies can be easily derived from the movements of the arm and trunk. Furthermore, they indicate that the use of different strategies can be manipulated through specific task constraints. By using these task constraints, training regimes can be designed that specifically focus on restitution, whereas without these task constraints people tend to train compensatory strategies (Michaelsen *et al.*, 2006)

1.3.2 **lower extremities**

For tasks in which only the paretic side is used in task execution, like in reaching and grasping, it is rather straight forward to make a distinction between compensatory strategies or restitution of original movement patterns. When the paretic and non paretic side contribute to task execution as in gait or balance control, distinguishing between the different recovery mechanisms gets a lot more complicated. For these tasks, not only in the affected leg a different control strategy can be present, but also in the non affected leg secondary adaptations in control strategies might occur to compensate for the impairments in the paretic leg. Hence, the actual task execution is a complex interplay between the exerted control in the paretic and non paretic leg.

Therefore, the definition of restitution and compensation needs to be refined for the lower extremities. Restitution during gait and balance can still be demonstrated by a return of the affected movement patterns on the paretic side to normal patterns. For example a return of appropriate knee flexion during the swing phase of walking (Daly *et al.*, 2006). Restitution might also be reflected in an increased contribution of the paretic leg in task execution. In the latter case, the task execution does not necessarily have to return to pre-lesion levels but the function of the paretic leg in accomplishing the task is (partly) recovered. From this perspective, compensation can be expressed in a greater contribution of the non paretic leg. This means that we need methods to evaluate the efficacy of the paretic and non paretic leg in light of the performance of the tasks to differentiate between restitution and compensation in the lower extremities.

walking

Walking consists of several subtasks that have to be accomplished successfully, like foot clearance during swing, body weight support and propulsion. Gait of stroke patients can improve by changes in the execution of each of these subtasks. As mentioned before, changes during the swing phase can be easily derived from the leg movements. However, recovery of subtasks in which both legs contribute and which mainly occur during the stance phase is a lot harder to assess. From the start of this decade, some methods have been developed that are used for this purpose. In the following paragraphs, these methods will be discussed in more detail, starting, with Induced Acceleration Analysis (Neptune *et al.*, 2001; Zajac *et al.*, 2003; Hof & Otten, 2005).

Induced Acceleration Analysis assesses the efficacy of the generated (coordinated) muscle output in the (non) paretic leg in subtasks, such as propulsion and body weight support. In this method the contribution of each separate muscle force/joint torque to the forward/backward acceleration and vertical acceleration of the Centre of Mass (CoM) is determined, which reflects the contribution to progression and body weight support respectively. Hitherto, this method has not yet been applied to its full potential to study gait in stroke. Only Higginson and colleagues (2006) used this method in a single stroke patient to assess the contribution of the different muscles to body weight support during midstance. Compared to healthy subjects, the paretic plantar flexors showed a decreased contribution to weight support and the knee flexors even an increased opposition of weight support. These changes were counteracted by an increased contribution of the knee extensors. These results indicate the potential of the method to increase our understanding of altered muscle coordination in the paretic and non paretic leg (Knutsson & Richards, 1979; Lamontagne *et al.*, 2000; Den Otter *et al.*, 2006a) and in doing so in the understanding of the recovery mechanisms. However, broad application of this method in pathological gait studies and in training evaluations has been limited as the method is based on sophisticated forward models of walking, which require long computational time even on today's computers.

These forward models are so extensive because they estimate the generated forces in the individual muscles during walking through optimization. In these optimization procedures, it is assumed that the muscle and joint properties of stroke patients are the same as those of healthy subjects. Different studies have indicated that several of these properties change as a consequence of stroke (see for overview Gracies, 2005), such as the force-length relationship of the muscle (Ada *et al.*, 2003) and the passive stiffness of the joints (Sinkjaer & Magnussen, 1994). Incorporating these changes into the models is possible and will probably result in different optimal activation patterns, with different estimated muscle forces. Consequently, the interpretation of the roles of the separate muscles will likely also be different. As long as no subject specific muscle properties are incorporated into the models, it might be better to restrict the analysis to the coordinated output of muscles around a joint, which is expressed in the joint torques. These joint torques can be calculated from measurements and do not require assumptions about muscle and joint properties or optimization procedures and as such decrease the likelihood of faulty interpretations.

Bowden and colleagues (2006) applied a less detailed but straight forward approach in determining the contribution of the paretic leg in progression during walking. They used the positive impulse of the ground reaction force in the forward/backward direction as a measure for the propulsion generated in each leg. Their results showed that the propulsion in the paretic leg as portion of the total propulsion showed a significant correlation with the severity of the hemiparesis. Interestingly, some patients with severe hemiparesis had normal walking velocities (>0.8 m/s) while their paretic leg only contributed les than 30% to the total propulsion, indicating that they were using compensatory strategies. By using the same data set, Balasubramaniam and colleagues (2007) provided additional evidence for compensatory strategies in gait. Subjects who generated the least paretic propulsion, walked with relatively longer paretic steps (forward distance between paretic foot placement and the non paretic foot position). This suggests that the propulsion of the non paretic leg might compensate for the decreased propulsion in the paretic leg and as such is responsible for the increased paretic step length.

balance

For balance control, no method exists to determine the contribution of the paretic and non paretic leg in task execution. In balance control, the task performance can be described as stabilization of the body CoM relative to the base of support (ground area enclosed by your feet). Different studies have assessed that stroke patients have a decreased ability to stabilize the CoM, whereas other studies showed that the regulating activity in the paretic leg during balance control is impaired (Ikai *et al.*, 2003; De Haart *et al.*, 2004). None of them directly related the generated activity to the stabilization of the CoM, which is required to determine the effectiveness of the generated activity in controlling balance.

Although not specifically developed for application in a rehabilitation setting, different approaches have been used to relate the generated regulating activity (muscle activation) to CoM movements (see for an overview Van der Kooij *et al.*, 2005). Many of these approaches falsely simplified the mutual relationship between muscular activity and CoM movements to a simple cause and effect relation. These approaches only took into account that muscular activity results in body movements and ignored that this movement on its turn influences the muscular activity through different feedback loops. Van der Kooij and colleagues (2005) argued that balance control needs to be perturbed and special identification techniques need to be used in order to assess the relation between muscle output and body movements. Their proposed method was originally developed for studying the stabilizing mechanism in healthy subjects. By extending their method it might be possible to objectively determine the contribution of the paretic and non paretic leg in balance control.

Another promising technique in determining the contribution of the separate legs in balance control is the aforementioned Induced Acceleration Analysis, with the requisite that it is based on joint torques. By using Induced Acceleration Analysis it is possible to elucidate how the joint torques can contribute to the forward/backward accelerations of the CoM which are required to control balance. From these accelerations, the efficacy of the generated joint torques in the paretic and non paretic leg to stabilizing the CoM can be deduced.

1.4 **restitution and compensation in recovery**

Different notions exist about whether therapy should emphasize the recovery of normal movements versus the teaching of compensatory strategies. Compensatory strategies can be regarded as an efficient way to maximize function. For example, many stroke patients suffer from a decreased ability to flex the knee, which limits them in achieving appropriate foot clearance during the swing phase. The lack of knee flexion can be substituted for using a circumduction strategy, which implies that hip abduction and pelvic rotation are used to achieve foot clearance. Although this strategy results in an increased walking ability, it might limit the potential recovery of more advanced skills such as walking stairs, as for walking stairs an increased knee flexion is required. This example adheres to the notion that compensatory strategies might improve function on the short term, however at the same time it might impede subsequent gains in motor function. On the other hand, restitution results in closer to normal movement patterns, yet this is no guarantee for better recovery of function than using a compensatory strategy (Huitema *et al.*, 2004; Kim & Eng, 2004).

Many of the methods presented in paragraph 1.3 to quantify restitution and compensation have recently been developed. These methods have only been applied in cross sectional studies to demonstrate the presence and efficiency of compensatory strategies in chronic stroke patients or are currently used to assess the effect of specific training programs (Bowden *et al.*, 2007). However, the quantification methods for the arm have recently been applied in a training study (Michaelsen *et al.*, 2006). This study investigated the effectiveness of a training protocol in which the use of normal movement patterns during reaching was encouraged by use of the previously mentioned trunk restraint (Michaelsen *et al.*, 2001). Subjects who received training with this restraint showed a decreased reliance on trunk movements and increased elbow extension whereas subjects who received the same training without the trunk restraint showed the opposite results. These differences were accompanied by greater improvements in motor function and impairment for the subjects trained with the trunk restraint. Remarkably, the trunk restraint was most effective for more severely affected stroke patients. Thus, by specifically emphasizing the use of normal movement patterns, restitution in chronic stroke patients can be promoted. Even if these subjects already had developed considerable compensation strategies.

Assessing the importance of restitution and compensation during the recovery of gait and balance was a main topic of the project "Herstel van lopen na een CVA" (Research program on recovery of gait following stroke, funded by ZonMW). In this project, two studies were conducted to determine whether the control strategy of the leg during gait returned to normal during the course of rehabilitation (Buurke, 2005; Den Otter *et al.*, 2006b). The control strategies were derived from measurements of the timing of the muscle activity of the major leg muscles. Functional recovery of gait was not accompanied by a reorganization in the temporal control of muscle activity neither in the paretic leg (Buurke, 2005; Den Otter *et al.*, 2006b) nor in the non-paretic leg (Buurke, 2005). Although these results do not provide conclusive evidence for restitution or compensation, they also do not exclude the importance of either mechanism in recovery. As these studies only regarded the timing of muscle activity, they cannot exclude that the amplitude of the muscle activity and as such the muscle force has changed over time. As at least something should have changed that accounts for the increased walking performance, it is likely that there are changes in the amplitude. Changes over time in the amplitude of separate muscles are hard to assess reliably, especially in stroke patients. However, changes in the combined coordinated output of the muscles of one leg can be derived from changes in the ground reaction force. On this notion the previously presented approach of Bowden and colleagues (2006) is based.

As part of the same ZonMW project, De Haart and colleagues (2004) assessed the recovery of balance control during quiet stance. They showed that balance control got more stable as expressed by a reduction of postural sway. The recovery of balance control was not associated with a restoration of the symmetry in the activity generated in the paretic and non paretic leg. Subjects sustained an increased reliance on activity generated in their non affected leg in controlling balance. Although these results point at compensatory mechanisms, the method they used might have obscured the occurrence of restitution. In their method, all the activity generated around the ankle was considered to stabilize the body and they did not directly relate the generated activity to the execution of the task. Consequently, badly coordinated activity or random activity in the paretic ankle is considered to be just as effective in stabilizing the body as well coordinated activity. Possibly, the amount of generated activity in the paretic leg did indeed not change, but the activity was better coordinated and as such was more efficient in postural stabilization, which would reflect restitution. These considerations indicate the need for more detailed methods to assess the efficacy of the paretic leg in controlling balance.

The aforementioned studies made a start to disentangle the recovery mechanisms. An increased understanding of these mechanisms is of great relevance for the design of new training strategies. The last decades there have been several studies investigating the effect of specific treatments on motor recovery. Recently, the effects of these studies have been summarized in a number of systematic reviews. These reviews identified the intensity of training as a key element in facilitating recovery during rehabilitation (Kwakkel *et al.*, 1997; Foley *et al.*, 2003; Kwakkel *et al.*, 2004b; Van Peppen *et al.*, 2004a; Teasell *et al.*, 2005). Another identified key element is the specificity of training. A systematic review (Van Peppen *et al.*, 2004a) of 152 randomized controlled trials and controlled clinical trials showed that training of functional tasks had the greatest efficacy. However, whether the focus during the repeated performance of the functional tasks should be on restitution or compensation is largely unknown and can be considered to be one of the next steps in optimizing rehabilitation strategies.

1.4.1 **the possible role of rehabilitation robotics**

Providing task specific and intensive therapy to stroke patients might not be as straightforward as it seems. Especially during the early stages of recovery the motor impairments impede the performance of even a single movement. A physical therapist can support the patient in making appropriate movements. However, guidance of repetitive movements places a high burden on the physical therapist. To relieve the therapist from this strenuous task, "gentle" robotic devices were introduced. These devices are attached to the limbs of the patients and can assist a person in the highly repetitive movements by exerting forces on the limb, much like the manual assistance provided by a therapist.

Robotic devices can play a crucial role in unraveling the recovery processes (Kwakkel *et al.*, 2007). These devices can be used to assess the most efficient strategy to regain function. This can be achieved by designing training protocols that only differ in the provided guidance, that is, guiding the movements in such a way that either the use of normal movement patterns or the use of alternative movement strategies is stressed. Not all rehabilitation devices are suitable for implementing this kind of training. The suitability is determined by the mechanical design of the robot and by the used control of the device. The mechanical design of the robot determines the number of Degrees of Freedom (DoF) that are assisted, left free or resisted by the robot and therefore prescribes which movements are possible in the device. The control of the device determines how movements are assisted and as such whether different movement strategies can be supported.

requirements for the mechanical design

A minimal requirement of a device to assess the importance of compensatory movement strategies is that the number of DoFs of the device should allow these alternative movement strategies. The number of assisted and free DoFs of the device should be larger than the number of DoFs of the task at hand, so the rehabilitation device provides redundancy. For instance, to make a planar reaching movement in the horizontal plane only shoulder and elbow flexion/extension are strictly necessary. However, if the device in addition to these two DoFs also allows shoulder abduction/adduction or the involvement of trunk movements different movement strategies can be utilized.

Most of the devices that are used for arm rehabilitation could allow alternative strategies, however the occurrence of these strategies is mostly prevented by additional movement

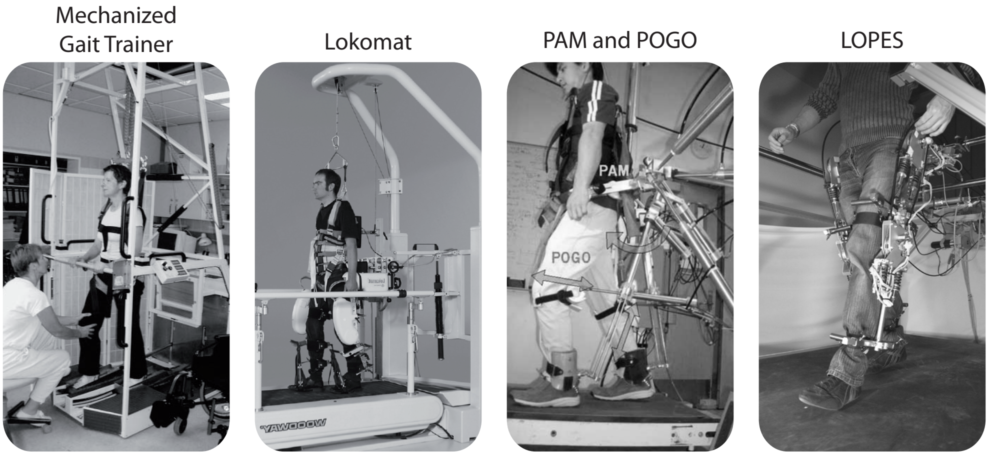

Figure 1.1. Overview of the robotic gait training devices.

constraints. The clinically evaluated robotic devices, such as the MIT-MANUS (or its commercial version InMotion2, Interactive Motion Technologies, Cambridge, USA) (Aisen *et al.*, 1997; Volpe *et al.*, 1999; Fasoli *et al.*, 2004; Daly *et al.*, 2005), the MIME (Burgar *et al.*, 2000; Lum *et al.*, 2006) and ACT3D (Sukal *et al.*, 2007) support reaching movement in 2D or in 3D by providing interaction forces at the lower arm or wrist. As the interaction is limited to the "end-effector" of the extremity, these devices are called end-effector robots. Although, at first glance it seems that only controlling the "end effector" of an extremity leaves room for compensatory movement strategies, the use of alternative movements is limited by imposing additional movement constraints. For example, during training of 2D reaching movements with the most widely used and tested robotic device MIT-MANUS, movements of the wrist are prevented by fixating the wrist in a fixed posture, abduction/adduction of the shoulder is prevented by a lower arm support and involvement of torso movements are constrained by a (five-point) seatbelt. Through these constraints the number of remaining DoFs at the different joints is equalized to the number of DoFs at the end-effector. This implies that guiding movements in a straight line will automatically impose a pattern of joint movements. Reinforcing this pattern comes down to reinforcing a normal movement strategy as healthy subjects make fairly straight movements. In short, arm rehabilitation robots mainly focus on relearning the original movement strategies, as redundant DoFs are constrained. By removing the additional movement constraints, patients could be left free in using compensatory strategies however in that case movements at the specific joints cannot be tracked or directly assisted anymore.

As this thesis mainly deals with lower extremities, the possibilities to allow and assist alternative movement strategies of the gait training devices will be discussed in more detail.

Currently, the mechanized gait trainer (Reha-Stim, Berlin, Germany)(Hesse *et al.*, 1999), the Autoambulator (Healthsouth, USA) and the market leading Lokomat (Hocoma AG , Volketswil, Switzerland) (Colombo *et al.*, 2000) are commercially available, whereas ALEX (Active Leg Exoskeleton)(Banala *et al.*, 2007), a combination of PAM (Pelvic Assist Manipulator) and POGO (Pneumatically Operated Gait Orthosis) (Aoyagi *et al.*, 2007) and LOPES (Lower Extremity Powered ExoSkeleton) (Veneman *et al.*, 2007) are under development at different research institutes (Figure 1.1). Of these devices the mechanized gait trainer is the only device that can be characterized as a pure end-effector robot. The other devices can be regarded as exoskeleton robots, in which the robot is attached to the controlled limb at several places, and the robot moves in parallel with the segments of the limb. In Table 1.1 an overview is provided of the assisted, free and constrained DoFs of the aforementioned devices. The mechanized gait trainer, Lokomat, ALEX and Autoambulator only allow and/or assist flexion/extension at the hip, knee and ankle and constrain all the movements out of the sagittal plane even though these movements are natural to human gait. As a consequence, alternative movement strategies are also not possible in these devices. For instance, the patients can only attain enough foot clearance through knee flexion and not by using a hip circumduction strategy as this would at least require the possibility to perform abduction. The recently at our department developed gait training device LOPES (an extensive description of the mechanical design can be found in the PhD thesis of Veneman (2007)) and the combination of the recently developed PAM and POGO provide (un)assisted abduction movement and different movement possibilities of the pelvis. These additional DoFs provide the required redundancy to allow the use of compensatory strategies.

In short, to allow compensatory strategies in a robotic device, the device should have more DoFs than strictly necessary for performing the task. In order to train these strategies, the movements should not only be possible but should also be supported. In other words, the involved DoFs should be actuated and not just free.

requirements for the controller

Apart from the mechanical design, there is another important consideration that determines whether a robot will allow alternative movement strategies and that is the specific control of the robot. Robots can be programmed in different ways to provide the mechanical guidance in the movements. The control approaches can broadly be subdivided in position control and impedance control. In position control a certain movement trajectory is enforced upon the patient. Position control hardly allows any deviation from these reference trajectories. In impedance control, not the positions but the forces are taken as the base of control.

Table 1.1. Overview of the actuated (A), free (F) and restrained (R) degrees of freedom of 6 different robotic devices. A dash indicates that the degree of freedom can be indirectly influenced by the provided assistance at the other DoFs

Joint/ Segment	Degrees of Freedom	Mechanized Gait Trainer	Lokomat Auto-ambulator Alex	PAM and POGO	LOPES
Pelvis	Vertical translation	-	F	A	F
	Horizontal translation	-	R	A	A
	Rotations	C/-	R	A	R
Hip	Flexion/extension	-	A	A	A
	Abduction/adduction	R	R	F	A
	Exo/endorotation	R	R	F	R
Knee	Flexion/extension	-	A	A	A
Ankle	Plantar/dorsiflexion	-	F	F	F
Foot	Vertical translation	A	-	-	-
	Forward/backward translation	A	-	-	-

However, reference position trajectories can still be used to determine the applied force. In this respect, an impedance controller can mimic the action of a regular spring. In a spring the deviation from the rest length, together with the stiffness of the spring determine the spring force. Likewise in impedance control, the regulated force can be calculated from the set stiffness and the deviation of the reference trajectory. By using small stiffness values, relatively small forces are applied, which allows subjects to deviate considerably from the reference trajectory and to influence movement patterns. On the contrary, when setting high values for the stiffness, large forces are applied in response to a deviation, which implies that the impedance controller merely acts as a position controller.

Both control regimes are frequently used in the control of rehabilitation robotics, however different names are often used to indicate a specific controller. Those names reflect the action required of the subject while training in the device. The passive approach indicates that the device is position controlled and that the subject is not required to generate activity. This approach is mostly used during gait training in the Lokomat (Colombo *et al.*, 2000), and always during training in the Autoambulator and Mechanized Gait Trainer (Hesse & Uhlenbrock, 2000; Hesse *et al.*, 2003). When using an “active assisted” approach, the device is impedance controlled and the subject generates his own movement and assistive forces are provided if the patient’s movements deviate from the optimal trajectory. The “active assisted” approach has been mainly used in the MIT-MANUS (Volpe *et al.*, 1999) and other arm rehabilitation devices (Kahn *et al.*, 2006) and has recently found its way to gait training devices like ALEX, combination of PAM and POGO and the LOPES. The active assisted approach also allows for adapting the amount of support based on the capabilities of the subject. When the subject recovers, the amount of support is decreased in such a way that the subject is only assisted as needed and is encouraged to improve further. These so called “assist-as-needed” algorithms have recently been shown to be more effective than a regular active assisted approach in training with the MIT-MANUS (Ferraro *et al.*, 2003; Hogan *et al.*, 2006).

The aforementioned approaches are based on following a certain reference trajectory and mainly differ in the amount of force used to force the patient to that trajectory. The exerted forces determine to what extent the generated movements can deviate from the reference trajectory and as such whether alternative movements are possible. In addition, the definition of the reference trajectory might even be more important in determining whether alternative strategies can be used or not. In end-effector robots the reference trajectory is generally defined as a straight path from the starting position to the target where different combinations of joint movements can result in this path as long as the robot allows kinematic redundancy. However, as mentioned before, this is not often the case. In exoskeleton robots, the reference trajectories are often defined in joint angles and are based on the trajectories of healthy subjects. The control of joint angles towards these “healthy trajectories” limits the flexibility in using alternative strategies. In fact, when using joint-based reference trajectories, compensatory movements could only be used if these movements are defined as the reference trajectories. However, defining and choosing between different compensatory reference trajectories, would be quite cumbersome as the optimal compensatory movement largely depends on the impairments of the subject. In this respect, defining subject-specific reference trajectories could form a solution. Recently, Aoyagi and colleagues (2007) implemented a teach and replay algorithm for controlling the pelvic motions during training with the PAM device. In this algorithm the reference trajectory is first recorded, while the subject is walking on the treadmill with the robot attached to the pelvis. During this phase, the robot is controlled in such a way that it does not actively assist the movement, but merely serves as a recording device. The necessary guidance is provided by a physical therapist, who can influence the movements of the subject and in doing so the eventual reference trajectory. In the subsequent training, this recorded trajectory is replayed what amounts to an endless repetition of the therapist's actions. This algorithm provides a way to incorporate alternative movement strategies in training with exoskeleton robots.

A completely different approach that also leaves room for compensatory strategies during training in an exoskeleton is to get away from controlling joint movements and to provide control at the level of subtasks. (Ekkelenkamp *et al.*, 2005; Ekkelenkamp *et al.*, 2007; Van Asseldonk *et al.*, 2007). This approach is used to control LOPES. As mentioned before, gait can be thought of as consisting of different subtasks that all have to be accomplished successfully to progress without falling (Pratt, 1995; Van der Kooij *et al.*, 2003). These subtasks include balance control, weight support, attaining appropriate foot clearance. Most of these subtasks can be accomplished in different ways that is by different joint movements. By executing control of a gait training device at subtask level, subjects are left free in the strategy they use to accomplish the subtask and will only receive assistance whenever the subtask is not executed satisfactorily. Therefore, the control of subtasks can be regarded as a typical example of an assist-as-needed algorithm, in which assistance is only applied when it is required. However, this brings forward another important aspect of this kind of control which is that no assistance should be applied if not needed. This means that if subjects need no assistance at all, they should be able to walk naturally in the device.

Teach and replay and selective support of subtask are promising techniques to incorporate compensatory strategies into robotic rehabilitation and are both examples of assist-as-needed algorithms. Still, they can only be used in this regard, if the DoFs of the robotic devices are redundant and if the necessary DoFs are actuated.

1.5 thesis objectives and goals

In recent years, there have been major advances in understanding how rehabilitative strategies can be used to promote recovery of motor function. Still, no consensus exists about which of the underlying recovery mechanisms, that is compensation or restitution, is the main contributor to motor recovery. Especially for the lower extremities evidence providing support for either of the mechanisms is very limited, as the function of the paretic and non paretic leg during tasks like balance control and gait can only be assessed by using biomechanical models and analysis. More insight in the responsible recovery mechanisms is needed to take the next step in designing rehabilitation strategies and that is how to treat the different limbs in order to achieve the most effective level of functioning.

Robotic devices are well suited to emphasize the use of specific strategies during functional tasks. However, using robotic device for this aim places additional demands on the mechanical design of the device and the provided assistance by the device.

The goal of this thesis is two fold

1. Develop and evaluate methods which can be used to distinguish between restitution and compensation in the recovery of function in the lower extremities of stroke survivors.
2. Provide a basis for the use of assist-as-needed algorithms which allow the flexibility to use different movement strategies during robot-aided training.

Essentially, this thesis has been built around these two different goals. Chapters 2 to 4 are related to the first goal and chapter 5 and 6 to the second.

Chapter 2 introduces a new method to determine the contribution of the ankle of the paretic and non paretic leg in balance control. By determining the separate contributions, we can deduce the functional role of the paretic leg in balance control. This method uses closed loop system identification techniques to relate the generated torques in each ankle to the balance responses of subjects to continuous horizontal surface perturbations. This method is evaluated in a group of chronic stroke survivors.

In chapter 3 a completely different approach is used to calculate the contribution of the different joints to stabilization of the human body in response to a surface perturbation. In contrast to the approach presented in chapter 2, this approach is not restricted to the contribution of the ankles and it uses transient instead of continuous perturbations. The method is based on an analytical approach to calculate the induced acceleration analysis. Although the described method can be used to gain a greater understanding of the contributions of and the coordination between the different joints of the paretic and non paretic leg to stabilization, this chapter contains data of one healthy subject to demonstrate the “proof of principle”.

Chapter 4 contains the evaluation of a new balance training that aims at improving the balance control in the paretic leg. The use of new technologies in the rehabilitation of stroke patients enhances the possibilities to elicit a functional response in the paretic leg. This training makes use of surface perturbations to specifically address the postural control in the paretic leg when the subject has to withstand perturbations. The method described in chapter 2 is used to evaluate if the training results in a greater contribution of the paretic leg which would reflect a restitution of function in the paretic leg. This evaluation is combined with an evaluation of the functional balance control and walking ability to get an overall picture of the efficacy of the training.

Chapter 5 addresses the question how assistance can be best provided during training. Different robotic devices have been developed to assist the movements of stroke patients during training. However, it is largely unknown what kind of assistance results in the optimal relearning of movements. Even in healthy subjects the effect of providing assistance during learning of a new task is yet unknown. We assessed the effect of different supportive forces, which differed in magnitude and direction during learning of a new task in healthy subjects. These guiding forces either attenuated or enlarged the errors made during the execution of the practice movements. The results of this study give us an indication about the usefulness of the different support modes in promoting relearning in stroke patients.

Chapter 6 investigates the basis of the implementation of assist-as-needed algorithms in LOPES. Assist-as-needed algorithms are used in robotic neurorehabilitation to optimize the cooperation of the patient by adapting the level of assistance to the capabilities of the patient. These algorithms require that ideally no assistance is applied if not needed and that as such normal walking should be possible for healthy subjects as long as essential movements are not constrained by the DoFs of the device. Still, the device will always interact in a certain extent with normal walking. In this study, the effect of walking in LOPES was assessed by comparing the joint movements and muscle activity while healthy subjects were walking with and without the robotic gait trainer. The results will show how the movement strategies of the subjects are affected when walking in the gait rehabilitation robot and as such how close we get to normal walking. Furthermore, the results will help us to identify the possible source(s) of the differences and will indicate how the design of the robot can be further improved.

1.6 references

Ada L, Canning CG & Low SL. (2003). Stroke patients have selective muscle weakness in shortened range. *Brain* 126: 724-731

Aisen ML, Krebs HI, Hogan N, McDowell F & Volpe BT. (1997). The effect of robot-assisted therapy and rehabilitative training on motor recovery following stroke. *Archives of Neurology* 54: 443-446.

Aoyagi D, Ichinose WE, Harkema SJ, Reinkensmeyer DJ & Bobrow JE. (2007). A robot and control algorithm that can synchronously assist in naturalistic motion during body-weight-supported gait training following neurologic injury. *IEEE Transactions on Neural Systems and Rehabilitation Engineering* 15: 387-400

Balasubramanian CK, Bowden MG, Neptune RR & Kautz SA. (2007). Relationship between step length asymmetry and walking performance in subjects with chronic hemiparesis. *Archives of Physical Medicine and Rehabilitation* 88: 43-49

Banala SK, Kulpe A & Agrawal SK. (2007). A Powered Leg Orthosis for Gait Rehabilitation of Motor-Impaired Patients. In *Proceedings of ICRA 2005 - IEEE International Conference on Robotics and Automation*: 4140-4145,. Roma, Italy.

Beer RF, Ellis MD, Holubar BG & Dewald JP. (2007). Impact of gravity loading on post-stroke reaching and its relationship to weakness. *Muscle & Nerve* 36: 242-250

Beer RF, Given JD & Dewald JP. (1999). Task-dependent weakness at the elbow in patients with hemiparesis. *Archives of Physical Medicine and Rehabilitation* 80: 766-772

Bowden MG, Balasubramanian CK, Neptune RR & Kautz SA. (2006). Anterior-posterior ground reaction forces as a measure of paretic leg contribution in hemiparetic walking. *Stroke* 37: 872-876

Bowden MG, Behrman AL & Kautz SA. (2007). Mechanisms of response to locomotor training post-stroke: systematic assessment of motor pattern restitution. In *International Society for Posture and Gait Research*, pp. 97. Burlington, Vermont, USA.

Burgar CG, Lum PS, Shor PC & Machiel Van der Loos HF. (2000). Development of robots for rehabilitation therapy: the Palo Alto VA/Stanford experience. *Journal of Rehabilitation Research and Development* 37: 663-673.

Buurke JH. (2005). *Walking after stroke: Co-ordination patterns & functional recovery*. PhD Thesis, Enschede.

Cirstea MC & Levin MF. (2000). Compensatory strategies for reaching in stroke. *Brain* 123: 940-953.

Colombo G, Joerg M, Schreier R & Dietz V. (2000). Treadmill training of paraplegic patients using a robotic orthosis. *Journal of Rehabilitation Research and Development* 37: 693-700.

Colombo R, Pisano F, Micera S, Mazzone A, Delconte C, Carrozza MC, Dario P & Minuco G. (2007). Assessing Mechanisms of Recovery During Robot-Aided Neurorehabilitation of the Upper Limb. *Neurorehabilitation and Neural Repair*

Daly JJ, Hogan N, Perepezko EM, Krebs HI, Rogers JM, Goyal KS, Dohring ME, Fredrickson E, Nethery J & Ruff RL. (2005). Response to upper-limb robotics and functional neuromuscular stimulation following stroke. *Journal of rehabilitation research and development* 42: 723-736

Daly JJ, Roenigk K, Holcomb J, Rogers JM, Butler K, Gansen J, McCabe J, Fredrickson E, Marsolais EB & Ruff RL. (2006). A randomized controlled trial of functional neuromuscular stimulation in chronic stroke subjects. *Stroke* 37: 172-178

De Haart M, Geurts AC, Huidekoper SC, Fasotti L & van Limbeek J. (2004). Recovery of standing balance in postacute stroke patients: a rehabilitation cohort study. *Archives of Physical Medicine and Rehabilitation* 85: 886-895

Den Otter AR, Geurts AC, Mulder T & Duysens J. (2006a). Abnormalities in the temporal patterning of lower extremity muscle activity in hemiparetic gait. *Gait & Posture*

Den Otter AR, Geurts AC, Mulder T & Duysens J. (2006b). Gait recovery is not associated with changes in the temporal patterning of muscle activity during treadmill walking in patients with post-stroke hemiparesis. *Clinical Neurophysiology* 117: 4-15

Dewald JP & Beer RF. (2001). Abnormal joint torque patterns in the paretic upper limb of subjects with hemiparesis. *Muscle & Nerve* 24: 273-283.

Dietz V & Sinkjaer T. (2007). Spastic movement disorder: impaired reflex function and altered muscle mechanics. *Lancet Neurology* 6: 725-733

Dipietro L, Krebs HI, Fasoli SE, Volpe BT, Stein J, Bever C & Hogan N. (2007). Changing motor synergies in chronic stroke. *Journal of Neurophysiology* 98: 757-768

Ekkelenkamp R, Veltink PH, Stramigioli S & van der Kooij H. (2007). Evaluation of a VMC for the selective support of gait functions using an exoskeleton. In *Proceedings of ICORR 2007 - IEEE International Conference on Rehabilitation Robotics*: 1055-1062. Noordwijk, The Netherlands.

Ekkelenkamp R, Veneman J & van der Kooij H. (2005). LOPES: selective control of gait functions during the gait rehabilitation of CVA patients. In *Proceedings of ICORR 2005 - IEEE International Conference on Rehabilitation Robotics*: 361-364. Chicago, Illinois, USA

Fasoli SE, Krebs HI, Stein J, Frontera WR, Hughes R & Hogan N. (2004). Robotic therapy for chronic motor impairments after stroke: Follow-up results. *Archives of Physical Medicine and Rehabilitation* 85: 1106-1111

Ferraro M, Palazzolo JJ, Krol J, Krebs HI, Hogan N & Volpe BT. (2003). Robot-aided sensorimotor arm training improves outcome in patients with chronic stroke. *Neurology* 61: 1604-1607

Foley NC, Teasell RW, Bhogal SK, Doherty T & Speechley MR. (2003). The efficacy of stroke rehabilitation: a qualitative review. *Topics in Stroke Rehabilitation* 10: 1-18

Gracies JM. (2005). Pathophysiology of spastic paresis. I: Paresis and soft tissue changes. *Muscle & Nerve* 31: 535-551

Hesse S, Schulte-Tigges G, Konrad M, Bardeleben A & Werner C. (2003). Robot-assisted arm trainer for the passive and active practice of bilateral forearm and wrist movements in hemiparetic subjects. *Archives of Physical Medicine and Rehabilitation* 84: 915-920

Hesse S & Uhlenbrock D. (2000). A mechanized gait trainer for restoration of gait. *Journal of Rehabilitation Research and Development* 37: 701-708.

Hesse S, Uhlenbrock D & Sarkodie-Gyan T. (1999). Gait pattern of severely disabled hemiparetic subjects on a new controlled gait trainer as compared to assisted treadmill walking with partial body weight support. *Clinical Rehabilitation* 13: 401-410.

Higginson JS, Zajac FE, Neptune RR, Kautz SA & Delp SL. (2006). Muscle contributions to support during gait in an individual with post-stroke hemiparesis. *Journal of Biomechanics* 39: 1769-1777

Hof AL & Otten E. (2005). Assessment of two-dimensional induced accelerations from measured kinematic and kinetic data. *Gait & Posture* 22: 182-188

Hogan N, Krebs HI, Rohrer B, Palazzolo JJ, Dipietro L, Fasoli SE, Stein J, Hughes R, Frontera WR, Lynch D & Volpe BT. (2006). Motions or muscles? Some behavioral factors underlying robotic assistance of motor recovery. *Journal of Rehabilitation Research and Development* 43: 605-618

Huitema RB, Hof AL, Mulder T, Brouwer WH, Dekker R & Postema K. (2004). Functional recovery of gait and joint kinematics after right hemispheric stroke. *Archives of Physical Medicine and Rehabilitation* 85: 1982-1988

Ikai T, Kamikubo T, Takehara I, Nishi M & Miyano S. (2003). Dynamic postural control in patients with hemiparesis. *American Journal of Physical Medicine & Rehabilitation* 82: 463-469

Kahn LE, Zygman ML, Rymer WZ & Reinkensmeyer DJ. (2006). Robot-assisted reaching exercise promotes arm movement recovery in chronic hemiparetic stroke: a randomized controlled pilot study. *Journal of Neuroengineering and Rehabilitation* 3: 12

Kim CM & Eng JJ. (2003). The relationship of lower-extremity muscle torque to locomotor performance in people with stroke. *Physical Therapy* 83: 49-57

Kim CM & Eng JJ. (2004). Magnitude and pattern of 3D kinematic and kinetic gait profiles in persons with stroke: relationship to walking speed. *Gait & Posture* 20: 140-146

Knutsson E & Richards C. (1979). Different types of disturbed motor control in gait of hemiparetic patients. *Brain* 102: 405-430

Kwakkel G, Kollen B & Lindeman E. (2004a). Understanding the pattern of functional recovery after stroke: facts and theories. *Restorative Neurology and Neuroscience* 22: 281-299

Kwakkel G, Kollen BJ & Krebs HI. (2007). Effects of Robot-Assisted Therapy on Upper Limb Recovery After Stroke: A Systematic Review. *Neurorehabilitation and Neural Repair*

Kwakkel G, van Peppen R, Wagenaar RC, Wood Dauphinee S, Richards C, Ashburn A, Miller K, Lincoln N, Partridge C, Wellwood I & Langhorne P. (2004b). Effects of augmented exercise therapy time after stroke: a meta-analysis. *Stroke* 35: 2529-2539

Kwakkel G, Wagenaar RC, Koelman TW, Lankhorst GJ & Koetsier JC. (1997). Effects of intensity of rehabilitation after stroke. A research synthesis. *Stroke* 28: 1550-1556

Lamontagne A, Richards CL & Malouin F. (2000). Coactivation during gait as an adaptive behavior after stroke. *Journal of Electromyography and Kinesiology* 10: 407-415

Lang CE, Wagner JM, Edwards DF, Sahrmann SA & Dromerick AW. (2006). Recovery of grasp versus reach in people with hemiparesis poststroke. *Neurorehabilitation and Neural Repair* 20: 444-454

Lennon S, Ashburn A & Baxter D. (2006). Gait outcome following outpatient physiotherapy based on the Bobath concept in people post stroke. *Disability and Rehabilitation* 28: 873-881

Levin MF, Michaelsen SM, Cirstea CM & Roby-Brami A. (2002). Use of the trunk for reaching targets placed within and beyond the reach in adult hemiparesis. *Experimental Brain Research* 143: 171-180

Lum PS, Burgar CG, Van der Loos M, Shor PC, Majmundar M & Yap R. (2006). MIME robotic device for upper-limb neurorehabilitation in subacute stroke subjects: A follow-up study. *Journal of Rehabilitation Research and Development* 43: 631-642

Michaelsen SM, Dannenbaum R & Levin MF. (2006). Task-specific training with trunk restraint on arm recovery in stroke: randomized control trial. *Stroke* 37: 186-192

Michaelsen SM, Jacobs S, Roby-Brami A & Levin MF. (2004). Compensation for distal impairments of grasping in adults with hemiparesis. *Experimental Brain Research* 157: 162-173

Michaelsen SM, Luta A, Roby-Brami A & Levin MF. (2001). Effect of trunk restraint on the recovery of reaching movements in hemiparetic patients. *Stroke* 32: 1875-1883

Mirbagheri MM, Settle K, Harvey R & Rymer WZ. (2007). Neuromuscular abnormalities associated with spasticity of upper extremity muscles in hemiparetic stroke. *Journal of Neurophysiology* 98: 629-637

Nadeau S, Gravel D, Arsenault AB & Bourbonnais D. (1999). Plantarflexor weakness as a limiting factor of gait speed in stroke subjects and the compensating role of hip flexors. *Clinical Biomechanics (Bristol, Avon)* 14: 125-135

Neptune RR, Kautz SA & Zajac FE. (2001). Contributions of the individual ankle plantar flexors to support, forward progression and swing initiation during walking. *Journal of Biomechanics* 34: 1387-1398

Pratt JE. (1995). Virtual model control of a biped walking robot. MIT, Cambridge.

Rohrer B, Fasoli S, Krebs HI, Hughes R, Volpe B, Frontera WR, Stein J & Hogan N. (2002). Movement smoothness changes during stroke recovery. *The Journal of Neuroscience* 22: 8297-8304.

Rohrer B, Fasoli S, Krebs HI, Volpe B, Frontera WR, Stein J & Hogan N. (2004). Submovements grow larger, fewer, and more blended during stroke recovery. *Motor Control* 8: 472-483

Sinkjaer T & Magnussen I. (1994). Passive, intrinsic and reflex-mediated stiffness in the ankle extensors of hemiparetic patients. *Brain* 117 (Pt 2): 355-363

Sukal TM, Ellis MD & Dewald JP. (2007). Shoulder abduction-induced reductions in reaching work area following hemiparetic stroke: neuroscientific implications. *Experimental Brain Research* 183: 215-223

Teasell R, Bitensky J, Salter K & Bayona NA. (2005). The role of timing and intensity of rehabilitation therapies. *Topics in Stroke Rehabilitation* 12: 46-57

Van Asseldonk EHF, Ekkelenkamp R, Veneman JF, van der Helm FCT & van der Kooij H. (2007). Selective control of a subtask of walking in a robotic gait trainer(LOPES). In *Proceedings of ICORR 2007 - IEEE International Conference on Rehabilitation Robotics:* 841-848. Noordwijk, The Netherlands.

Van der Kooij H, Jacobs R, Koopman B & Van der Helm F. (2003). An alternative approach to synthesizing bipedal walking. *Biological Cybernetics* 88: 46-59

Van der Kooij H, Van Asseldonk E & Van der Helm FC. (2005). Comparison of different methods to identify and quantify balance control. *Journal of Neuroscience Methods* 145: 175-203

Van Peppen RP, Hendriks HJ, Van Meeteren NL, Helders PJ & Kwakkel G. (2007). The development of a clinical practice stroke guideline for physiotherapists in The Netherlands: a systematic review of available evidence. *Disability and Rehabilitation* 29: 767-783

Van Peppen RP, Kwakkel G, Wood-Dauphinee S, Hendriks HJ, Van der Wees PJ & Dekker J. (2004a). The impact of physical therapy on functional outcomes after stroke: what's the evidence? *Clinical Rehabilitation* 18: 833-862

Van Peppen RPS, Kwakkel G, Harmelink BC, Kollen BJ, Hobbelen JSM, Buurke JH, Halfens J, Wagenborg L, Vogel MJ, Berns M, Van Klaveren R, Hendriks HJM & Dekker J. (2004b). Clinical Practice Guideline in Physiotherpay-magement of patients with Stroke [in Dutch: KNGF-richtlijn Beroerte]. *Nederlands Tijdschrift voor Fysiotherapie* 114: 1-78

Veneman JF, Kruidhof R, Hekman EEG, Ekkelenkamp R, Van Asseldonk EHF & Van der Kooij H. (2007). Design and Evaluation of the LOPES Exoskeleton Robot for Interactive Gait Rehabilitation. *IEEE Transactions on Neural Systems and Rehabilitation Engineering* 15: 379-386

Veneman JF, (2007). *Design and evaluation of the gait rehabilitation robot LOPES.* PhD Thesis, Enschede.

Volpe BT, Krebs HI, Hogan N, Edelsteinn L, Diels CM & Aisen ML. (1999). Robot training enhanced motor outcome in patients with stroke maintained over 3 years. *Neurology* 53: 1874-1876

Wagner JM, Lang CE, Sahrmann SA, Hu Q, Bastian AJ, Edwards DF & Dromerick AW. (2006). Relationships between sensorimotor impairments and reaching deficits in acute hemiparesis. *Neurorehabilitation and Neural Repair* 20: 406-416

Wu CY, Chen CL, Tang SF, Lin KC & Huang YY. (2007). Kinematic and clinical analyses of upper-extremity movements after constraint-induced movement therapy in patients with stroke: a randomized controlled trial. *Archives of Physical Medicine and Rehabilitation* 88: 964-970

Zajac FE, Neptune RR & Kautz SA. (2003). Biomechanics and muscle coordination of human walking: part II: lessons from dynamical simulations and clinical implications. *Gait & Posture* 17: 1-17

chapter 2

disentangling the contribution of the paretic and non-paretic ankle to balance control in stroke patients

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2.1 abstract

During stroke recovery, restitution of the paretic ankle and compensation in the non-paretic ankle may contribute to improved balance maintenance. We examine a new approach to disentangle these recovery mechanisms by objectively quantifying the contribution of each ankle to balance maintenance. Eight chronic hemiparetic patients were included. Balance responses were elicited by continuous random platform movements. We measured body sway and ground reaction forces below each foot to calculate corrective ankle torques in each leg. These measurements yielded the Frequency Response Function (FRF) of the stabilizing mechanisms, which expresses the amount and timing of the generated corrective torque in response to sway at the specified frequencies. The FRFs were used to calculate the relative contribution of the paretic and non-paretic ankle to the total amount of generated corrective torque to correct sway. All patients showed a clear asymmetry in the balance contribution in favor of the non-paretic ankle. Paretic balance contribution was significantly smaller than the contribution of the paretic leg to weight bearing, and did not show a clear relation with the contribution to weight bearing. In contrast, a group of healthy subjects instructed to distribute their weight asymmetrically showed a one-on-one relation between the contribution to weight bearing and to balance. We conclude that the presented approach objectively quantifies the contribution of each ankle to balance maintenance. Application of this method in longitudinal surveys of balance rehabilitation makes it possible to disentangle the different recovery mechanisms. Such insights will be critical for the development and evaluation of rehabilitation strategies.

2.2 introduction

The mechanisms underlying clinical recovery during rehabilitation of stroke remain unclear. This is particularly true when both the paretic and non-paretic side contribute to task execution, e.g., in postural control where improved performance following acute stroke is usually ascribed to recovery ("restitution") of the paretic leg (Geurts *et al.*, 2005). However, alternative explanations are possible, because secondary adaptations ("compensation") of the non-paretic leg can occur that compensate for impairments in the paretic leg (Kirker *et al.*, 2000; Garland *et al.*, 2003; de Haart *et al.*, 2004). Restitution and compensation are not mutually exclusive, and can concur in a single patient.

Separation of both processes has clinical implications, e.g., for designing rehabilitation strategies (Bloem *et al.*, 2001). For postural control, this requires quantifying the contribution of each leg to overall task performance. Theoretically, improved task performance that is accompanied by greater contributions of the paretic leg should index that the paretic leg has (partially) restored its efficacy, and vice versa. Changes in muscle activity (Kirker *et al.*, 2000; Garland *et al.*, 2003) may occur in one or both legs, but by themselves, such changes do not indicate their efficacy in improving postural control. To deduce whether compensation and/or restitution take place, it is essential to quantify how the observed changes contribute to postural control.

Such an approach requests experiments where the individual efficacy of both legs can be reliably separated. This can be achieved by determining the contribution of generated activity (i.e., EMG, joint torque) to stabilization of the center of mass (CoM). Previously used methods falsely considered balance control as an open-loop system, and assumed a causality between generated activity and CoM movements (for an overview, see Van der Kooij *et al.*(2005)), i.e., cross correlation was used to determine the time lag between EMG activity and CoM. In reality, however, balance control is a closed-loop system where the generated activity acts on the body mechanics resulting in segment movements. This in turn influences the generated activity by different feedback loops, involving the muscle spindles, Golgi tendon organs and portions of the central nervous system.

Advanced closed-loop system identification techniques with a well-defined external perturbation signal are needed to determine the causal relations in a closed loop. Van der Kooij et al (2005) treated balance as a closed-loop system and presented a method to relate CoM movements to generated ankle torque, based on a biomechanical model of balance control and knowledge of system identification.

Here, we study the merits of this method in determining the individual contribution of the paretic and non-paretic ankle to postural control in chronic stroke patients. We elaborated upon the original approach and incorporated two ankles instead of one, so the stabilizing contribution of corrective torque in the paretic and non-paretic ankle could be determined separately. The contribution of both ankles to balance maintenance is expressed as a fraction, much like weight distribution. We predicted that an asymmetry in the balance contribution is not a mere reflection of an asymmetry in weight distribution. To test this hypothesis, we assessed weight distribution and compared it with the specific balance contribution of each ankle. In a control experiment, we compared the weight distribution of both legs to the balance contribution when healthy age-matched subjects adopted an asymmetrical weight distribution. This allowed us to examine if weight distribution and balance contribution are directly related or, as we predicted, at least partially independent measures.

2.3 materials and methods

2.3.1 subjects

Eight patients with hemiparesis secondary to a single and first ever unilateral stroke in the territory of the anterior or middle cerebral arteries participated. Subjects were at least 1 year post-stroke, had no musculoskeletal or neurological diseases in addition to stroke, were able to remain standing without help or support for at least 90 s and were able to adequately comprehend our instructions. As our aim was to evaluate the feasibility of the proposed method, we included a large range of functional abilities in our group of subjects. To characterize the patient population (Table 2.1), all patients were scored on the Berg Balance Scale (Berg, 1989), the Timed Balance Test (Bohannon *et al.*, 1984), the Timed Up and Go test (Podsiadlo & Richardson, 1991), the Functional Ambulation Categories (Holden *et al.*, 1984) and the Motricity Index of the paretic leg (Collin & Wade, 1990). Six healthy age-matched subjects (6 men, mean age: 61.4 (standard deviation: 3.4)) participated in the control experiment.

The experiment was approved by the local medical ethical committee and conformed to the principles of the Declaration of Helsinki. All subjects gave their written informed consent prior to the start of the experiment.

2.3.2 theoretical background

The method is developed based on a commonly used model of postural control (Fitzpatrick *et al.*, 1996; Peterka, 2002) (see Figure 2.1). In this model, upright stance of the human body is stabilized by corrective torques acting on the human body around the ankles. This model also assumes that the human body acts like an inverted pendulum, which is inherently unstable.

When a spontaneous small deviation of upright stance occurs, gravity induces an acceleration of the body even further away from upright stance. Internal or external perturbations can further destabilize posture. In order to maintain upright stance, a corrective torque is generated by posturally stabilizing mechanisms that are based on the "sensed" sway angle (i.e., the deviation of

Table 2.1. Summary of subject characteristics.

Patient	Sex	Age	Time since stroke	Affected hemisphere	Stroke type	AFO	BBS	TBT	TUG	FAC	MI	DA
1	M	43	23	Left	hemorrhagic	Yes	52	4	27.1	5	39	8
2	M	56	36	Right	ischemic	Yes	54	4	11.8	5	44	7
3	M	62	25	Left	ischemic	No	56	4	18.6	5	44	7
4	M	59	29	Right	ischemic	No	51	3	9.1	5	75	7
5	M	62	69	Right	ischemic	Yes	44	3	24.2	4	37	4
6	M	68	52	Left	hemorrhagic	Yes	34	3	46.6	4	37	7
7	M	59	61	Left	hemorrhagic	No	56	4	13.7	5	72	8
8	M	70	44	Right	ischemic	Yes	45	3	23.4	5	64	6

Time since stroke is expressed in months. AFO, indicates the use of an Ankle Foot Orthoses. The Berg Balance Scale (BBS) assesses static and dynamic balance on a scale from 0 to 56. The timed balance test (TBT) assesses the balance performance in 5 different standing postures on a scale from 0 to 5 . The Timed Up and Go test (TUG) assesses the mobility and expresses the time [s] needed to stand up from a chair, walk 3 m, turn and return to sitting position. The Functional Ambulation Categories (FAC) assesses the independency while walking on a scale from 0 to 5. The Motricity Index (MI) of the lower extremity assesses the motor impairment of the lower extremity on a scale from 0 to 100. DA indicates the amplitude of the disturbance signal.

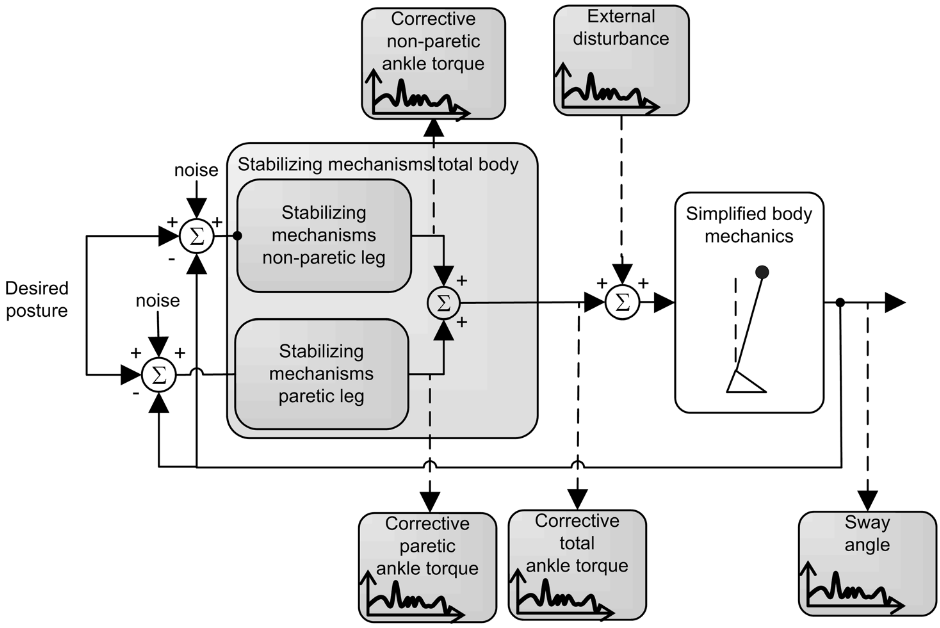

Figure 2.1. A model of postural control of a stroke patient. The simplified body mechanics represent the dynamics of an inverted pendulum with the total corrective torque as input and the sway angle as output. The sway angle is compared to the desired posture, which is upright stance. The deviation of upright stance with the addition of noise serves as input for the stabilizing mechanisms. The stabilizing mechanisms represent the dynamics of the combination of active and passive feedback pathways of the concerned body(part) and generate a torque to correct for the deviation of upright stance. Noise represents the unknown noise on the reference signal, errors in the perceived sway angle and insufficiencies of the model. The graph icons denote which signals are measured or can be calculated from measured signals in this study.

the human body with respect to upright stance). These stabilizing mechanisms use an active and a passive feedback pathway that all generate part of the corrective torque (Peterka, 2002). In the active feedback pathway, the central nervous system (CNS) perceives the sway by means of sensory signals from the visual, vestibular and proprioceptive/somatosensory systems and subsequently generates efferent signals that ultimately result in muscle contractions and corrective torques. In the passive feedback pathway, visco-elastic properties of the muscles and tissue around the ankle joint instantly generate a corrective torque in response to sway. In this study, we will not discriminate between these different pathways and only consider the combined effect of the different pathways.

Previous balance control studies (Kiemel *et al.*, 2002; Peterka, 2002) that used the inverted pendulum model only considered the net corrective torque from both ankles together. This assumption is justified in healthy subjects where both ankles contribute equally to the generation of corrective torques. However, in stroke patients, we expect that the sensory and motor impairments of the hemiparetic side will negatively affect its ability to generate a corrective torque. In this case, the non-paretic ankle should compensate for this deficiency by generating more corrective torque. In order to quantify the contribution of the paretic and non-paretic leg in maintaining upright stance, we split up the total stabilizing mechanism into two parallel stabilizing mechanisms: the paretic and non-paretic stabilizing mechanisms (see Figure 2.1). Each of these stabilizing mechanisms consists of the sensory and motor system of the concerned leg and the part of the CNS that processes sensory signals and generates efferent signals to the motor system. Both stabilizing mechanisms produce a corrective torque, which sum up to produce the required corrective torque to overcome the effect of gravity and internal or external bodily perturbations. In terms of the inverted pendulum model, the sum of both torques stabilizes the one inverted pendulum representing the total human body.

The contribution of the paretic and non-paretic leg to postural control can be determined by quantifying the proportion of the total corrective torque generated by each separate leg in response to a deviation of upright posture. This requires that the relation between sway and the corrective torques is established, what comes down to identifying the content of the stabilizing mechanisms. The stabilizing mechanisms contain position and velocity feedback and neural time delays and can therefore be regarded as dynamical systems. This means that the amplification and timing of their response, the corrective torque, depend on the frequency content of their input, the sway angle. In order to determine the dynamical characteristics of the stabilizing mechanisms, the FRF can be estimated by using the joint input–output approach (Ljung, 1999; Van der Kooij *et al.*, 2005). The joint input–output approach can only be applied when external mechanical and/or sensory perturbations are applied to the balancing human and not during quiet stance. In this study, horizontal platform movements in the forward–backward directions were used to perturb the balance of stroke patients, which made it possible to determine the contribution of paretic and non-paretic leg in balance control.

2.3.3 apparatus and recording

The participants stood with their arms folded in front of their chest or hanging at their sides on a force plate, embedded in a computer-controlled 6 degrees of freedom motion platform (Caren, Motek, Amsterdam, The Netherlands). All subjects were allowed to wear comfortable shoes and, if needed, patients were allowed to wear an Ankle Foot Orthosis (AFO) (see Table 1). The feet were placed with their medial sides and heels against a fixed foot frame resulting in a separation of the medial sides of the heels of ±20 cm and an outward rotation of the feet of 9° with respect to the sagittal midline. The participants faced a visual scene (4.80×3.70 m at a distance of 3.20 m) consisting of a light grey background with a life-sized dummy of a human just in front of it.

The force plate consisted of four 6 degrees of freedom force sensors (ATI-Mini45-SI-580-20, supplier: Schunk, Arnhem, NL), mounted in a 2×2 configuration on an aluminum plate. Each sensor was covered with a rectangular aluminum plate with a dimension of 15×17.5 cm. In between both left force sensors and both right force sensors, an aluminum plate was placed to assure the desired stance width. The foot frame ensured that each foot was placed solely on the cover of either the 2 right force sensors or the 2 left force sensors. Forces and torques of each sensor were sampled at a frequency of 360 Hz.

Reflective spherical markers were attached to the heel, big toe, malleolus, tibia, knee and femur of both legs, as well as the sacrum, head and both shoulders. Moreover, a cluster of 3 markers was attached to the anterior superior iliac spine of both legs and 3 markers were attached to the platform. The movement trajectories of the markers were recorded at a sampling rate of 120 Hz by means of a three-dimensional passive registration system (Vicon Oxford Metrics, Oxford, UK), consisting of six infrared sensitive cameras and a control unit. Body length and weight were measured for each subject. Subjects wore a safety harness suspended from the ceiling, which prevented the subject from falling, but did not constrain the movements necessary to maintain balance or provide any support or orientation information.

2.3.4 **procedures**

In the experiment, subjects stood on the motion platform and were instructed to "maintain their balance without moving their feet" while continuous random platform movements were applied in the forward–backward direction. The reason for using continuous random movements is explained in the next section. Before data recording, the perturbation was presented to the patients in order to let them get acquainted to the test conditions and to determine the amplitude of the disturbance during the recordings. For the stroke patients, the zero-to-peak amplitude of the first practice trial was set at 3 cm. Subsequently, the amplitude was raised between the practice trials with steps of 1 or 2 cm as long as the patient and an accompanying physical therapist were confident that the patient could withstand this amplitude for 90 s.The final amplitude was used in three trials of 90 s each in which the response of the subjects was measured. In the control experiment, the amplitude during the habituation trials and recording trials was set to 10 cm. The healthy age matched subjects were instructed to adopt an asymmetrical standing posture during the perturbation trials, mimicking the asymmetrical posture of stroke patients, by shifting their weight to their right leg, while maintaining contact with the floor with their left leg.

The disturbance trials were preceded by a static trial in which subjects were instructed to adopt their normal posture for 90 s. To prevent fatigue, patients were given ample time for a standing or seated rest between the trials. The time interval between trials was thus dictated by the subject and ranged between 1 and 2 min. In the rare occasion that subjects had to make a small step to prevent falling, the trial was repeated immediately.

2.3.5 **disturbance signal**

The random continuous platform movements consisted of a multisine signal. A multisine was used because it is periodic, contains power only at the desired frequencies and is unpredictable, preventing a contribution of anticipation to the postural response. To prevent leakage, each harmonic should fit exactly an integer number of times in the multisine signal (Pintelon & Schoukens, 2001). This is assured by only including harmonics with frequencies equal to an integer multiple of the frequency resolution, which is defined as the inverse of the period of the multisine. The multisine had a period of 34.13 s (equal to 2^{12} =4096 samples at a sample rate of 120 Hz) and contained 80 frequencies ranging from 2/34.13=0.06 Hz to 81/34.13=2.37 Hz.

The power at each frequency decreased logarithmically with logarithmically increasing frequency. The ratio of the power at the lowest frequency and at the highest frequency is 6308. The variance of the signal was optimized with respect to its amplitude by crest optimization (Pintelon and Schoukens, 2001). In crest optimization, the phases of the sines are optimized such that the variance is maximal at a given amplitude. The perturbation trials lasted 90 s, in which the above-described multisine was repeated time after time. The disturbance signal was scaled to provide the different zero-to-peak amplitudes mentioned in Table 2.1.

2.3.6 **data analysis**

static trial

The measured forces and torques from the four sensors were resampled to a frequency of 120 Hz and subsequently digitally low pass filtered with a second order recursive Butterworth filter with a cut-off frequency of 6 Hz. The forces and torques from both sensors below each foot were used to calculate the vertical and horizontal forces below each foot and their point of application, the Center of Pressure (CoP). From the vertical forces, the static weight-bearing asymmetry is calculated by dividing the average vertical force below the paretic foot by the average of the sum of the vertical forces below both feet.

dynamic trials

The signals denoted with a graph icon in Figure 2.1 were calculated from the measured data. From the recorded marker positions, the position of the CoM was calculated, by first calculating the separate rotations and positions of the body segments and subsequently determining the CoM as the weight sum of separate segment positions (Koopman *et al.*, 1995). The length of the pendulum (l_{CoM}) was determined from the static trial as the average distance in the sagittal plane from the ankle to the CoM. The sway angle was calculated from the length of the pendulum and the horizontal distance from the CoM to the mean position of the ankles.

In the dynamic trial, the measured forces and torques from the four sensors were first resampled to a frequency of 120 Hz and subsequently low pass filtered as in the static trial. Subsequently, the forces and torques were corrected for the inertia and mass of the top cover (procedures according to Preuss and Fung (2004)). The corrected forces and torques from both sensors below each foot were used to calculate the vertical and horizontal forces below each foot and their point of application, the CoP. The ankle torque in the paretic ($T_{ank\text{-}P}$) and the non-paretic leg ($T_{ank\text{-}NP}$) were calculated with inverse dynamics (Koopman *et al.*, 1995). The total generated corrective torque ($T_{ank\text{-}T}$) is the sum of the ankle torques at both ankles.

The vertical forces below both feet were also used to calculate the dynamic weight-bearing (DW) asymmetry in the same way as done in the calculation of the static weight-bearing asymmetry.

Platform movements by themselves are not perturbing human balance, but the accelerations of the platforms are. The magnitude of the perturbation further depends on the mass of the subject minus the mass of the feet (m_{CoM}) and the distance from the ankles to the CoM, in other words the length of the pendulum (Van der Kooij *et al.*, 2005).

$$d_{ext} = -m_{CoM} \cdot l_{CoM} \cdot \ddot{x}_{sb} \qquad (1)$$

where d_{ext} indicates the external disturbance torque, and $\ddot{x}_{sb}$ the forward-backward acceleration of the support base.

The perturbation signal of 90 s consisted of approximately 2.5 cycles of the multisine in each of the three trials. From each trial's perturbation (d_{ext}) and response (sway, $T_{ank\text{-}T}$, $T_{ank\text{-}P}$ and $T_{ank\text{-}NP}$) trajectories 2 synchronized, succeeding, complete cycles (2^{12} = 4096 samples ≈ 34.13 s) were extracted each starting with the sample corresponding to a defined point in the perturbation signal, the rest of the trial was discarded. Subsequently, the perturbation and response cycles were decomposed in their sinusoidal component parts by using Discrete Fourier Transform

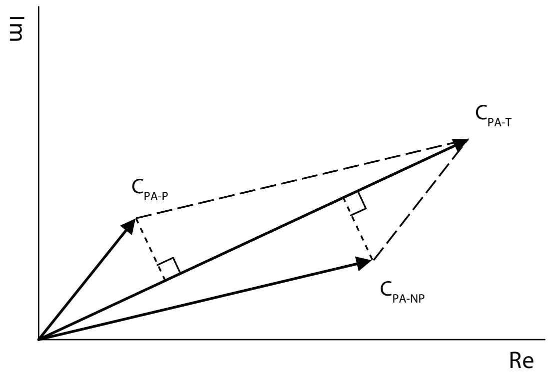

Figure 2.2. Graphical representation of stabilizing mechanisms of the paretic and the non-paretic leg and the total body. The real and imaginary part of each stabilizing mechanism is plotted as a vector on the real (Re) and imaginary (Im) axis. The angle of each of the vectors with the horizontal indicates the phase, while the length of the vector indicates the gain. The stabilizing mechanism of the total body is the vectorial sum of the stabilizing mechanisms of both legs. The contribution of each leg to the total gain can be obtained by a projection of the corresponding vector on the vector of the total stabilizing mechanisms.

(DFT). The DFT was applied to each of the 6 (2 cycles×3 trials) cycles of 34.13 s. The DFT was calculated at the 80 different frequencies of the perturbation signal.

From the obtained Fourier coefficients, the following Cross Spectral Densities (CSD) were calculated and averaged over all the cycles: CSD of the external disturbance torque with each of the ankle torques and CSD of the external disturbance torquewith the body sway. As stated in the Introduction, the stabilizing mechanisms are embedded in a closed-loop configuration, see Figure 2.1, therefore a joint input–output approach (Ljung, 1999; Van der Kooij *et al.*, 2005) was adopted to obtain the estimate of FRF of the stabilizing mechanisms. The FRF of the total stabilizing mechanisms (C_{PA-T}), the paretic stabilizing mechanisms (C_{PA-P}) and the non paretic stabilizing mechanisms (C_{PA-NP}) were estimated by dividing the averaged CSD of the external disturbance torque and the body sway with the CSD of the external disturbance torque and the corresponding corrective ankle torque.

The frequency dependency of the characteristics of the stabilizing mechanisms can be shown by calculating the gain and phase for each frequency from the estimated FRFs. The gain represents the ratio of the amplitude of the corrective torque to the amplitude of the sway angle and the phase gives the relative timing of the corrective torque with respect to the sway angle. The calculation of the gain and phase of the stabilizing mechanisms is illustrated in Figure 2.2. FRFs are complex numbers and each complex number can be depicted as a vector in the imaginary plane. The length of the vector and the angle with the vertical determine the gain and phase, respectively.

The gain of the stabilizing mechanisms of the total body indicates the total amount of corrective ankle torque generated in response to a deviation of upright stance at each frequency. In order to determine the contribution of each ankle to the generation of the total corrective torque, the contribution of the gain and phase of each leg to the gain of the total body was calculated.

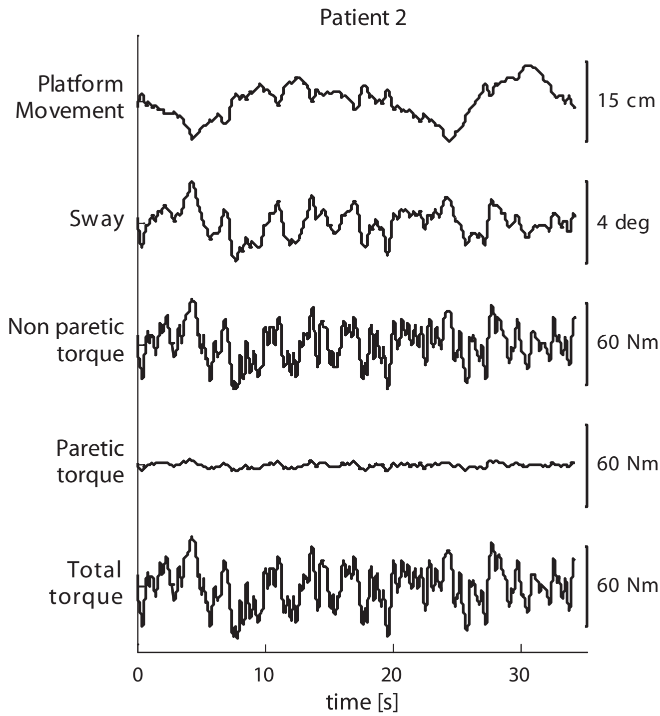

Figure 2.3. Example time series of a complete cycle of the multisine platform movement, the sway angle and the corrective ankle torque of the paretic leg, the non-paretic leg and the total body of a representative stroke patient.

In Figure 2.2, a graphical representation of the calculations is given. The total stabilizing mechanisms ($C_{PA\text{-}T}$) is the sum of the stabilizing mechanisms of the paretic ($C_{PA\text{-}P}$) and non-paretic leg ($C_{PA\text{-}NP}$). In the imaginary plane, $C_{PA\text{-}T}$ is the vectorial sum of $C_{PA\text{-}P}$ and $C_{PA\text{-}NP}$ (see Figure 2.2). The contribution of the paretic and non-paretic stabilizing mechanisms to the gain of the total stabilizing mechanisms was determined by projecting the vector of both stabilizing mechanisms on the vector of the total stabilizing mechanism. Division of the result by the total gain led to the contribution of each of the stabilizing mechanisms to the total expressed as a proportion.

In the foregoing, the contribution was calculated for each frequency, by averaging the contribution over the different frequencies, the Dynamic Balance Contribution (DBC) of the paretic and non-paretic ankle to the overall balance is determined. The data analysis for the control experiment was essentially the same, except that the calculations were performed for the right and left ankle instead of the paretic and non-paretic ankle.

statistical analysis

A paired t-test was used to compare the paretic dynamic balance contribution with the non-paretic dynamic balance contribution, the paretic dynamic balance contribution with the paretic dynamic weight distribution and the paretic static weight distribution with the paretic dynamic weight distribution.

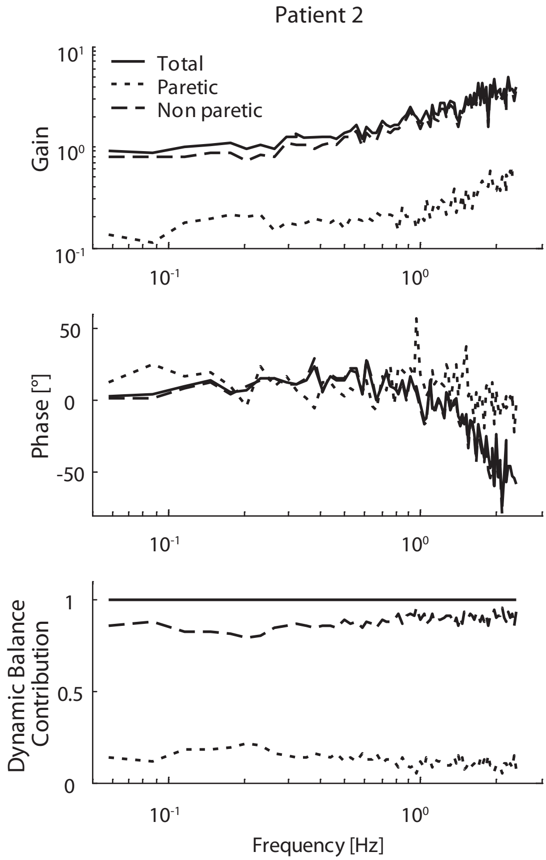

Figure 2.4. Stabilizing mechanisms of a representative stroke survivor. From the top to the bottom panel the gain, phase and the contribution to the total gain of the total, the paretic and non-paretic stabilizing mechanisms are depicted. The gains are scaled to the gravitational stiffness.

2.4 **results**

2.4.1 **time series**

The inertia of the body mass resulted in body sway that was directed opposite to the platform movements (Figure 2.3). This sway had to be counteracted by the generated corrective torques. However, the direct relation between sway and the generated corrective ankle torques was less clear. Inspection of the corrective torques showed that the fluctuations of the torques in the paretic leg were much smaller than the fluctuations in the non-paretic leg, indicating a lower sensitivity of the paretic torques to deviations of upright stance compared to the non-paretic torques. A further result is that almost all fluctuations in total torque of the stroke patients resulted from fluctuations in the non-paretic leg.

Higher frequency components were more obvious in the torque profiles than in the sway. This could indicate that the stabilizing mechanisms were more sensitive to higher frequencies and consequently amplified the higher frequencies to a greater degree than the lower ones.

Although the time series plots give an indication of the ratio of joint torques and sway and its dependency on frequencies, the relationship was covered by the closed loop nature of balance control. This urges further analyses of the data in the frequency domain by means of frequency response functions.

2.4.2 frequency response functions

In Figure 2.4, the FRFs of the stabilizing mechanisms of a typical stroke patient are depicted. The gain of the controller was scaled by the gravitational stiffness ($m_{CoM}l_{CoM}g$). Hence, a gain of one indicates that, in response to deviations of erect stance, the generated torque by the stabilizing mechanisms was exactly equal to the torque resulting from gravity. A gain greater than one indicates that the generated corrective torque was larger than the gravitational torque at the particular frequency. Consequently, the stabilizing mechanisms "push" the body back to erect posture. A gain of less than one points to a smaller corrective torque than the gravitational torque. In this case, the stabilizing mechanisms failed to generate enough torque to compensate for the gravitational torque. Whenever this occurs for the total stabilizing mechanisms, this would theoretically mean that the body falls over.

All gain profiles showed more or less the same pattern, with a gain that is lowest for the smallest frequencies and that gradually increases with higher frequencies. The gain of the $C_{PA\text{-}T}$ was smaller than one for the lowest frequencies for this patient and all the other patients. In line with the expectations from the time series, the gain of the $C_{PA\text{-}NP}$ was higher for all frequencies than the gain of the $C_{PA\text{-}P}$. The phase profile of the non-paretic leg resembled the one of the total body over the entire frequency range. The phase profile of the paretic leg diverged from these two profiles. For the lower frequencies, the phase was mostly greater than zero which means that the generated corrective torque preceded deviations of erect stance. The phase dropped below zero for the higher frequencies, indicating that the generated corrective torque was delayed with respect to the deviation of upright stance.

In the lower graph, the contribution of the stabilizing mechanisms of each separate leg to the gain of $C_{PA\text{-}T}$ is depicted. The large difference in magnitude of the gains of the paretic ankle and non-paretic ankle reflected on their contribution, the paretic ankle showed a much smaller contribution for all frequencies than the non-paretic ankle.

2.4.3 dynamic balance contribution

In order to determine the overall contribution of the paretic and non-paretic ankle to balance maintenance, the contributions were averaged over all frequencies. The resulting values are shown in Figure 2.5, together with the static and dynamic weight distribution. The weight distribution was not significantly different when the patients were standing on the motionless platform or when they were withstanding the platform perturbations (P=0.917), indicating that patients maintained their normal posture during the perturbation trials. The patients showed a broad range of asymmetry in their weight distribution, from patients greatly relying on their non-paretic leg for weight support (patient 5 and 6) to patients standing almost symmetrically (patient 7), or even supporting more weight on their paretic leg (patient 4). Regardless of the amount of asymmetry in weight distribution, all patients except one (patient 4) showed a clear asymmetry in balance contribution (range: 0.11–0.45 DBC of the paretic leg). The dynamic balance contribution of the paretic ankle was significantly smaller ($P<0.001$) than the contribution of the non-paretic ankle.

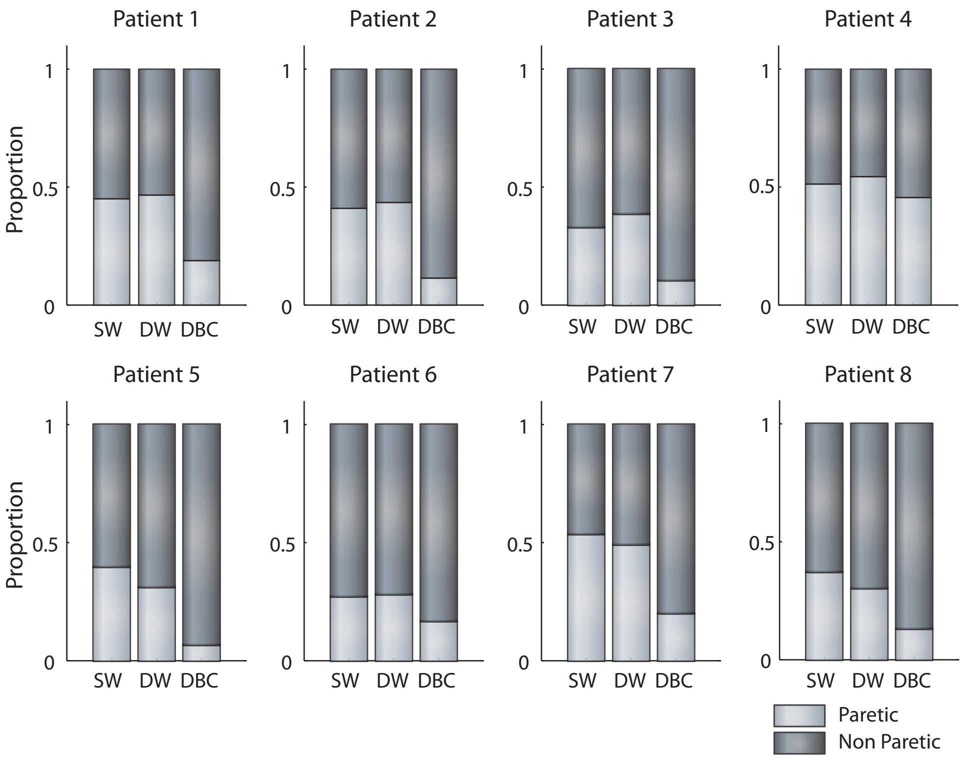

Figure 2.5. Weight distribution during quiet stance (SW), weight distribution during the perturbation trials (DW), and the dynamic contribution to balance maintenance (DBC) for the paretic and the non-paretic leg for 8 patients. All values are expressed as proportions and consequently sum up to 1.

For a closer look at the relation between the weight distribution and the contribution to balance, the paretic DBC is plotted versus the paretic DW for every patient (squares in Figure 2.6). All the squares are underneath the dashed line, which indicates that for all patients the paretic leg contributed less to balance than to weight ($P < 0.001$). Apart from this difference, all squares are clustered together, with one clear outlier (patient 4). In order to further examine the relationship between weight bearing and balance contribution, we performed a regression analysis, by fitting a regression line through all combinations. The initial results were largely dependent on the outlier (square in the upper right corner of Figure 2.6). This was expressed in a high value for the Cook's distance for this particular case, indicating that the residuals of all cases would change substantially if this case was excluded from the calculation and in a high value for the Leverage value, indicating that the outlier had a great influence on the fit of the regression line. Exclusion of the outlier from the calculation of the regression line, resulted in non-significant regression ($P=0.24$). The regression line had a slope of 0.32, an intercept of 0.01 and explained 26.3% of the variance. This means that there was no clear relation between the contribution of the paretic leg to weight bearing and balance.

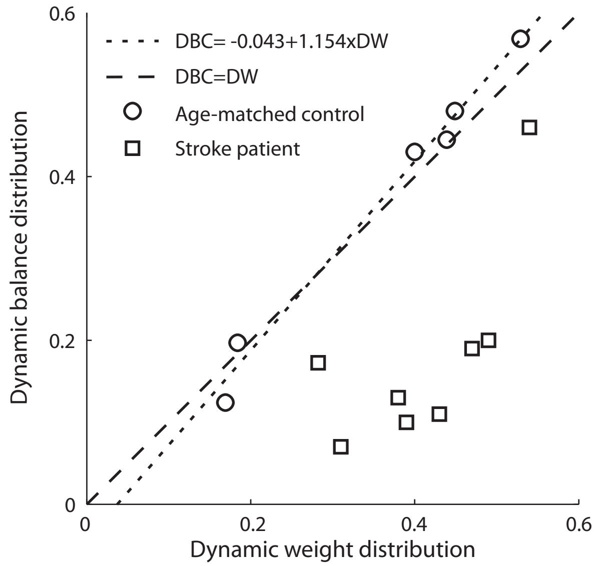

Figure 2.6. Dynamic weight distribution (DW) versus dynamic balance contribution (DBC) for the paretic leg of stroke patients (squares) and the left leg of healthy age-matched subjects while obtaining an asymmetrical stance (circles). The dashed line indicates where the balance contribution is exactly equal to the weight bearing. The dotted line is the fitted regression line through the combinations of the dynamic balance contribution and weight distribution of the left leg of the healthy subjects.

2.4.4 **control experiment**

In the control experiment, healthy subjects adopted an asymmetrical posture by shifting weight to their right leg and unloading their left leg. As they were left free in adopting this posture, they unloaded the left leg in different amounts (DW on the left leg ranged from 0.17 to 0.53). The shift in DW was always accompanied by a similar shift in the DBC, the circles in Figure 2.6 were clustered around the dashed line which indicated DW equal to DBC (see Figure 2.6). A regression analysis confirmed the tight relationship between weight bearing and balance contribution. Fitting of a regression line through all the combinations of balance contribution and weight bearing during asymmetrical stance resulted in a regression line ($P<0.001$) with a slope close to 1 (1.15) and an intercept very close to 0 (−0.04) which explained 97.1% of the variance. All this showed that in healthy subjects the weight bearing and balance contribution were very closely interrelated, a relationship that was absent in stroke patients.

2.5 **discussion**

The goal of this paper was to assess a new method to determine the contribution of the paretic and non-paretic ankle to balance maintenance. For this purpose, it was necessary to relate the overall balance performance, in terms of the sway angle, to the torque generated in each of the two ankles. To achieve this, we had to consider the total balance control mechanisms and applied – for the first time – theoretical models and methods (Van der Kooij *et al.*, 2005) to identify such stabilizing mechanisms in a patient population. The results showed that, even in patients with a fairly symmetrical weight distribution, a clear asymmetry in balance contribution could be observed in favor of the non-paretic leg. This asymmetry was even more pronounced in patients with an asymmetrical weight distribution. Overall, the paretic ankle made a significantly smaller balance contribution than the non-paretic ankle.

One key finding was that the contribution of the paretic leg to balance was much smaller than its contribution to weight bearing and neither showed a clear relation with the contribution to weight bearing. By inference, this implies that the contribution of the paretic leg to balance is not a mere reflection of the weight distribution. This could have major consequences for the emergent balance training therapy using biofeedback, where the major focus is on restoration of symmetrical weight bearing in order to improve balance function (Shumway-Cook *et al.*, 1988). Biofeedback training involves providing stroke patients with visual feedback of their CoP and instructing them to hold the CoP as close as possible to a midline in order to reinstall symmetry of stance. It is implicitly assumed that restoration of a more symmetrical weight distribution will lead to a greater involvement of the paretic limb in postural control and thereby to a better postural control. At first sight, this assumption seems supported by observations made in healthy subjects, because our control experiment showed that the weight distribution and contribution to balance seem directly interrelated in healthy subjects. On the contrary, the results for the stroke patients challenge this assumption and show that under pathological conditions, weight distribution cannot be equated with actual contributions of individual legs to overall balance control. This could perhaps explain why a recent review failed to find any additional effects of biofeedback training on balance control – other than a greater stance symmetry – over and above conventional training strategies (Barclay-Goddard *et al.*, 2004).

The finding that the contribution of the paretic leg to balance control was smaller than its contribution to weight bearing is according to our expectations, as weight bearing is a rather static process, while balance control is highly dynamic. Therefore, the net clinical effect of the motor and sensory deficits might particularly come to light during dynamic balance control. Yet, the effect of sensory impairments on balance control has not been determined unambiguously. Niam and colleagues (1999) showed that impaired ankle proprioception led to greater postural sway in stroke patients. These results were not confirmed by two recent studies (Bonan *et al.*, 2004; Marigold *et al.*, 2004) in which the Sensory Organization Test was used to assess the ability to use sensory information from different modalities (visual, proprioceptive and vestibular) separately, as well as the ability to integrate this afferent information from the different modalities. Absence of ankle proprioceptive information on its own led to no (Bonan *et al.*, 2004) or only a relatively small decrease (Marigold *et al.*, 2004) in balance control compared to age-matched controls. Furthermore, these studies showed that there was no correlation between motor control in the paretic leg (Bonan *et al.*, 2004) or muscle strength in the paretic and non-paretic leg on the one hand, and balance control during quiet stance on the other hand. Although one can question whether the measure of balance control used in these studies (which was calculated from the sway range) was able to capture the complex task of balance control, these results indicate that the degree of sensory and motor deficits in the paretic leg do not determine the impairments in postural control.

However, in the Sensory Organization Test, the balance control was assessed on the level of the total body and not on the level of the separate legs. The relation between the impairments in the paretic leg and balance control might get obscured by compensation of the non-paretic leg, i.e., patients use their paretic leg as a crutch to bear weight on, while they perform balance control tasks with their non-paretic leg. The degree of compensation by the non-paretic leg, and the remaining function in the paretic leg, can be quantified with the dynamic balance contribution approach illustrated in the present study. Therefore, we expect a higher correlation between clinically rated motor and sensory impairments and the dynamic balance contribution. These relations will have to be assessed in future studies with larger sample sizes, as the patients in this study lacked the necessary variability on the scores of the administered tests.

Assessment of the DBC in a longitudinal study of balance rehabilitation in patients with stroke makes it possible to disentangle restitution, all adaptations that occur in the paretic leg during rehabilitation, from compensation, all adaptations that occur in the non-paretic leg. Repeated assessment will result in the dynamic balance contribution of the paretic and non-paretic ankle as a function of time. An improvement of the balance performance accompanied by an increase of the paretic DBC indicates restitution of function in the paretic leg, while an accompanying increase of the non-paretic DBC indicates compensation in the non-paretic leg. Many prior studies focused on balance impairment in stroke patients, typically using force plates to characterize postural control (Rode *et al.*, 1997; Niam *et al.*, 1999; Nardone *et al.*, 2001; Laufer *et al.*, 2003). The weight distribution and postural stability obtained from the CoP movements were generally used to assess the impairments in balance control. These measures appropriately assess to what degree balance is impaired in stroke patients and can therefore be used to evaluate balance improvements during rehabilitation. However, these methods cannot discriminate between restitution of the paretic leg versus compensation of the non-paretic leg. For this purpose, at least the generated activity has to be measured separately in each leg. Only relatively few studies of stroke patients separately analyzed the forces or CoP movements below each foot to estimate the stabilizing activity generated at both ankles, either during quiet stance (Mizrahi *et al.*, 1989; de Haart *et al.*, 2004) or in response to external perturbations (Chaudhuri & Aruin, 2000; Ikai *et al.*, 2003). Only one study repeatedly assessed the regulatory activity of each leg and their contribution to balance during a rehabilitation program (de Haart *et al.*, 2004). In this study, the Kinetic Regulation Asymmetry Quotient (KRAQ) was used to distinguish between restitution of the paretic leg versus compensation of the non-paretic leg. The KRAQ expresses the regulatory activity of the non-paretic leg relative to the paretic leg, using the quotient of the Root Mean Square (RMS) of the CoP velocity of each leg.

The DBC has some advantages compared to the KRAQ. First, the DBC relates the regulatory activity to the actual sway movements, while the KRAQ merely calculates the proportional regulatory activity of both leg, without links to actual balance performance. Second, the DBC is based on a biomechanical model of balance control and proper identification techniques (Van der Kooij *et al.*, 2005), while the KRAQ (which it is determined from the RMS of the CoP velocity) lacks such a theoretical basis. The RMS of CoP velocity is used for its good reliability and its good sensitivity to task manipulations or consequences of disease pathology. However, it cannot directly be interpreted within the scope of a balance model. Third, the DBC is calculated using the generated corrective ankle torques, which can be regarded as the combined effect of all muscles and connective tissues crossing the ankles. In contrast, the KRAQ is based solely on the CoP. Especially during quiet stance, a fairly good approximation of the generated ankle torque can be obtained by multiplying the CoP position relative to the ankle by the vertical ground reaction forces, which equals the weight born on a given leg. Consequently, when the weight is distributed asymmetrically, as is often the case for stroke patients, KRAQ actually underestimates the asymmetry in regulatory activity based on the generated corrective torques in both ankles. Finally, the DBC is assessed using external perturbations, while the KRAQ is assessed during quiet stance. Although use of external perturbations is more cumbersome than simple quiet stance, it is the only way to estimate the stabilizing mechanisms and distinguish those from the destabilizing mechanisms (Van der Kooij *et al.*, 2005). Estimating the stabilizing mechanisms is not possible during quiet stance, because no distinction can be made between the generated torques that stabilize and destabilize the body (Van der Kooij *et al.*, 2005). This is also true for the CoP movements, as fluctuations in CoP partially result from noise or arbitrary movements not aimed to restore balance. Such noise and arbitrary movements are probably more prominently present in the paretic leg, leading to less asymmetry in the KRAQ.

Several aspects of this study merit further discussion. We only assessed the contribution of the ankle to balance maintenance. Although the ankle plays a crucial role in different strategies to maintain balance (Alexandrov *et al.*, 2005), other joint likely contributed to the movements of the center of mass and as such to the maintenance of balance (Runge *et al.*, 1999; Park *et al.*, 2004). This might explain why the subjects were able to maintain their balance even when the gain of the total stabilizing mechanisms was smaller than one, indicating that less corrective ankle torque was generated than required to compensate for the gravitational torque. A gain lower than one could also be due to measurement inaccuracies, like errors in the estimation of the position of the Center of Mass. In future research, we will focus on the possibility to extend the method to incorporate the contributions of the other joints of the legs.

In this study, we determined the disturbance amplitude individually for each patient (see Table 2.1). The amplitude was set at the maximal amplitude that patients could comfortably withstand, because the destabilizing external torques should be as large as possible compared to the internally produced destabilizing torques to increase the reliability of the estimate of the stabilizing mechanisms (Van der Kooij *et al.*, 2005). Apart from large amplitude, it is crucially important that patients responded consistently through the different cycles. To facilitate this, we tried to maximize the patients' confidence on the platform by incorporating their judgement in the determination of the perturbation amplitude. So the amplitude of the perturbation signal was determined by their physical capabilities and the patients' natural confidence in their abilities. Although this leads to a relatively large range of disturbance amplitudes between the patients (see Table 2.1), this probably will not have affected the results because we studied the relative contribution of the legs and we did not quantify the balance in an absolute sense.

Five of our patients wore an Ankle Foot Orthosis (AFO) during the study, and this may have affected their postural performance. Wearing an AFO can affect several aspects of sway in patients with an acute stroke (Wang *et al.*, 2005). However, two studies showed that wearing an AFO does not affect postural sway and weight distribution during quiet stance in chronic stroke patients (Chen *et al.*, 1999; Wang *et al.*, 2005). The effect of the AFO was previously assessed during quiet stance or weight shifting tasks, so we cannot automatically extrapolate these findings to sway responses to external perturbations, as was used in the present study. Under such circumstances, the influence of an AFO may be accentuated because sway induced by external perturbations is much greater than sway during quiet stance. An AFO is designed to provide extra stability to the ankle. As a result of the stiffness of the AFO, a corrective torque is generated in response to sway. This torque adds up to the actively and passively generated corrective torque of the paretic ankle, together resisting the induced sway. For the present study, we did not investigate the stabilizing effect of the AFO, because our primary interest was to evaluate overall asymmetries in the stabilizing contribution of the paretic and non-paretic ankle. For the patients, wearing an AFO, the found asymmetries cannot be fully subscribed to the stroke, as the AFO could have influenced the results. The effect of an AFO on the dynamic balance contribution of the paretic leg will be a topic for future studies.

The patient population consisted of a relatively small group of eight patients. Yet, the effects were very clear and consistent, which underscores the strength of the observed effects and provides "proof of principle" of the dynamic balance approach. Relating the DBC scores with the clinical scores, in particular the Berg Balance Scale (BBS), might reveal some other interesting aspects. The BBS assesses functional balance in stroke patients. Five out of eight subjects in our study scored close to maximum on the BBS (>50 of 56). The lack of variability in the BBS scores discouraged us from doing a correlation analysis. However, the BBS results contained some interesting implications for the DBC scores. These implications can be revealed by looking at patients with extreme scores on the BBS. Four of the five patients with a close to maximum BBS score showed a rather low contribution of the paretic leg to balance control (pareticDBC<0.2). This implies that restitution of function in the paretic leg is not a prerequisite for a good functional balance. Apparently, the non-paretic leg can compensate for the loss of function in the paretic leg in such a way that it is still possible to have a good functional balance. However, the results for patient 6 show that compensation of the non-paretic leg does not always lead to a good functional balance. Although this patient had a paretic DBC in the range of the four patients mentioned before, his BBS score was the lowest of all patients. These results raise the question of what determines whether compensation leads to a restitution of function.As far as we are aware, not much is known about this topic.

In this study, the dynamic balance contribution was evaluated in a group of chronic stroke patients. The same method has great potential value for other patient populations, which are also characterized by a clear asymmetry in the impairments of both legs. Possible target populations include, among others, patients with Parkinson's disease (Bloem *et al.*, 2004), patients with a prosthetic leg (Buckley *et al.*, 2002). For example, for the latter population, it would be interesting to know whether the prosthetic leg contributes to balance control at all, or whether it merely serves for weight bearing. Furthermore, the new technique could help to clarify whether the introduction of more advanced prosthetic legs leads to a greater contribution to balance of the prosthetic leg. Such insights will also be critical for the development and evaluation of individually tailored rehabilitation strategies, where restitution of the affected leg should be facilitated and compensation in the unaffected leg be trained and promoted.

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2.7 **references**

Alexandrov A, Frolov A, Horak F, Carlson-Kuhta P & Park S. (2005). Feedback equilibrium control during human standing. *Biological Cybernetics* 93: 309-322

Barclay-Goddard R, Stevenson T, Poluha W, Moffatt ME & Taback SP. (2004). Force platform feedback for standing balance training after stroke. *Cochrane Database of Systematic Reviews* CD004129

Berg K. (1989). Measuring balance in the elderly: A Review. *Physiotherapy Canada* 41: 304-311

Bloem BR, Hausdorff JM, Visser JE & Giladi N. (2004). Falls and freezing of gait in Parkinson's disease: a review of two interconnected, episodic phenomena. *Movement Disorders* 19: 871-884

Bloem BR, Visser JE & Carpenter MG. (2001). Compensatory postural strategies in patients with longstanding balance impairment. In *Balance at all times*, ed. Van der Burg JCE, Fong FB, Hijl MIJ, Huys R, Pijnappels M & Post AA, pp. 51-73. Utrecht:Digital Printing Partners BV, Utrecht.

Bohannon RW, Larkin PA, Cook AC, Gear J & Singer J. (1984). Decrease in timed balance test scores with aging. *Physical Therapy* 64: 1067-1070

Bonan IV, Colle FM, Guichard JP, Vicaut E, Eisenfisz M, Tran Ba Huy P & Yelnik AP. (2004). Reliance on visual information after stroke. Part I: Balance on dynamic posturography. *Archives of Physical Medicine and Rehabilitation* 85: 268-273

Buckley JG, O'Driscoll D & Bennett SJ. (2002). Postural sway and active balance performance in highly active lower-limb amputees. *American Journal of Physical Medicine & Rehabilitation* 81: 13-20

Chaudhuri S & Aruin AS. (2000). The effect of shoe lifts on static and dynamic postural control in individuals with hemiparesis. *Archives of Physical Medicine and Rehabilitation* 81: 1498-1503

Chen CL, Yeung KT, Wang CH, Chu HT & Yeh CY. (1999). Anterior ankle-foot orthosis effects on postural stability in hemiplegic patients. *Archives of Physical Medicine and Rehabilitation* 80: 1587-1592

Collin C & Wade D. (1990). Assessing motor impairment after stroke: a pilot reliability study. *Journal of Neurology, Neurosurgery & Psychiatry* 53: 576-579

de Haart M, Geurts AC, Huidekoper SC, Fasotti L & van Limbeek J. (2004). Recovery of standing balance in postacute stroke patients: a rehabilitation cohort study. *Archives of Physical Medicine and Rehabilitation* 85: 886-895

Fitzpatrick R, Burke D & Gandevia SC. (1996). Loop gain of reflexes controlling human standing measured with the use of postural and vestibular disturbances. *Journal of Neurophysiology* 76: 3994-4008

Garland SJ, Willems DA, Ivanova TD & Miller KJ. (2003). Recovery of standing balance and functional mobility after stroke. *Archives of Physical Medicine and Rehabilitation* 84: 1753-1759

Geurts ACH, De Haart M, Van Nes IJW & Duysens J. (2005). A review of standing balance recovery from stroke. *Gait & Posture* 22: 267-281

Holden MK, Gill KM, Magliozzi MR, Nathan J & Piehl-Baker L. (1984). Clinical gait assessment in the neurologically impaired. Reliability and meaningfulness. *Physical Therapy* 64: 35-40

Ikai T, Kamikubo T, Takehara I, Nishi M & Miyano S. (2003). Dynamic postural control in patients with hemiparesis. *American Journal of Physical Medicine & Rehabilitation* 82: 463-469

Kirker SG, Jenner JR, Simpson DS & Wing AM. (2000). Changing patterns of postural hip muscle activity during recovery from stroke. *Clinical Rehabilitation* 14: 618-626.

Kiemel T, Kelvin SO, Jeka JJ (2002) Multisensory fusion and stochastic structure of postural sway. *Biological Cybernetics* 87: 262–277.

Koopman B, Grootenboer HJ & de Jongh HJ. (1995). An inverse dynamics model for the analysis, reconstruction and prediction of bipedal walking. *Journal of Biomechanics* 28: 1369-1376

Laufer Y, Sivan D, Schwarzmann R & Sprecher E. (2003). Standing balance and functional recovery of patients with right and left hemiparesis in the early stages of rehabilitation. *Neurorehabilitation and Neural Repair* 17: 207-213

Ljung L. (1999). *System identification: Theory for the user*. Prentice-Hall, Upper Saddle River.

Marigold DS, Eng JJ, Tokuno CD & Donnelly CA. (2004). Contribution of muscle strength and integration of afferent input to postural instability in persons with stroke. *Neurorehabilitation and Neural Repair* 18: 222-229

Mizrahi J, Solzi P, Ring H & Nisell R. (1989). Postural stability in stroke patients: vectorial expression of asymmetry, sway activity and relative sequence of reactive forces. *Medical and Biological Engineering and Computing* 27: 181-190

Nardone A, Galante M, Lucas B & Schieppati M. (2001). Stance control is not affected by paresis and reflex hyperexcitability: the case of spastic patients. *Journal of Neurology, Neurosurgery & Psychiatry* 70: 635-643.

Niam S, Cheung W, Sullivan PE, Kent S & Gu X. (1999). Balance and physical impairments after stroke. *Archives of Physical Medicine and Rehabilitation* 80: 1227-1233

Park S, Horak FB & Kuo AD. (2004). Postural feedback responses scale with biomechanical constraints in human standing. *Experimental Brain Research* 154: 417-427

Peterka RJ. (2002). Sensorimotor integration in human postural control. *Journal of Neurophysiology* 88: 1097-1118.

Pintelon R & Schoukens J. (2001). *System identification: a frequency domain approach*. IEEE Press, New York, USA.

Podsiadlo D & Richardson S. (1991). The timed “Up & Go”: a test of basic functional mobility for frail elderly persons. *Journal of the American Geriatrics Society* 39: 142-148

Preuss R & Fung J. (2004). A simple method to estimate force plate inertial components in a moving surface. *Journal of Biomechanics* 37: 1177-1180

Rode G, Tiliket C & Boisson D. (1997). Predominance of postural imbalance in left hemiparetic patients. *Scandinavian Journal of Rehabilitation Medicine* 29: 11-16.

Runge CF, Shupert CL, Horak FB & Zajac FE. (1999). Ankle and hip postural strategies defined by joint torques. *Gait & Posture* 10: 161-170

Shumway-Cook A, Anson D & Haller S. (1988). Postural sway biofeedback: its effect on reestablishing stance stability in hemiplegic patients. *Archives of Physical Medicine and Rehabilitation* 69: 395-400.

Van der Kooij H, Van Asseldonk E & Van der Helm FC. (2005). Comparison of different methods to identify and quantify balance control. *Journal of Neuroscience Methods* 145: 175-203

Wang RY, Yen L, Lee CC, Lin PY, Wang MF & Yang YR. (2005). Effects of an ankle-foot orthosis on balance performance in patients with hemiparesis of different durations. *Clinical Rehabilitation* 19: 37-44

chapter 3

use of induced acceleration to quantify the (de)stabilization effect of external and internal forces on postural responses

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3.1 abstract

Due to the mechanical coupling between the body segments, it is impossible to see with the naked eye the causes of body movements and understand the interaction between movements of different body parts. The goal of this paper is to investigate the use of induced acceleration analysis to reveal the causes of body movements. We derive the analytical equations to calculate induced accelerations and evaluate its potential to study human postural responses to support-surface translations. We measured the kinematic and kinetic responses of a subject to sudden forward and backward translations of a moving platform. The kinematic and kinetics served as input to the induced acceleration analyses. The induced accelerations showed explicitly that the platform acceleration and deceleration contributed to the destabilization and restabilization of standing balance, respectively. Furthermore, the joint torques, coriolis and centrifugal forces caused by swinging of the arms, contributed positively to stabilization of the Centre of Mass. It is concluded that induced acceleration analyses is a valuable tool in understanding balance responses to different kinds of perturbations and may help to identify the causes of movement in different pathologies.

3.2 introduction

Postural control is often assessed by using sudden transient support base movements. The activity generated to restore balance is quantified by variables deduced from force plates, motion analysis or Electromyography (EMG) recordings. The EMG recordings indicate which muscles show an increased activity in response to the perturbations and are therefore likely involved in counteracting the perturbations. However the contribution of a muscle to the balance restoration cannot be revealed solely by its EMG recording. Despite recent advances (Bogey *et al.*, 2005; Staudenmann *et al.*, 2006) in determining the relations between EMG activity and muscle forces, the validity of these relations in multi-joint movements has yet to be examined. Furthermore, the relative contribution of generated joint torques to balance control is currently unknown.

Besides internally generated activity other components may also contribute to balance restoration. McIlroy and Maki (McIlroy & Maki, 1994) were the first to point out that the deceleration of a moving platform could be considered as a second perturbation. The platform deceleration acts as a force on the human body which is opposite to the perturbing force of the acceleration, and therefore, may assist the human body to restore equilibrium. The potential of the deceleration phase to facilitate balance recovery was evident in a number of studies. Runge and colleagues (Runge *et al.*, 1999) showed that the reversal of the Centre of Mass (CoM) movement towards equilibrium coincided with the deceleration of the platform movement for backward perturbations with different velocities. By using responses to perturbations with different velocities, Bothner and Jensen(Bothner & Jensen, 2001) showed that an increased platform velocity was accompanied by an increased stabilization originating from the platform deceleration which even compensated for the decreased stabilization originating from the muscles. In a recent study, Carpenter and colleagues (Carpenter *et al.*, 2005) manipulated the delay between the equal initial acceleration and the onset of the deceleration phase (short delay: 0.1 vs long delay: 2 s). They clearly showed the stabilizing effect of the deceleration; angular displacements of the trunk were decreased in translations with a short, compared to long acceleration-deceleration interval. Furthermore they showed the ability of the Central Nervous System (CNS) to adapt its response to an initial acceleration on the basis of expectations about the forthcoming deceleration. These different studies clearly show the potential of a platform deceleration in restoring equilibrium; however the extent to which the platform deceleration contributes to movements of the joint and the CoM is currently unknown.

Another possible contributor to stabilization of the human body in response to external perturbations is arm movement. Arm movements could aid in restoring balance when the movements occur in the direction opposite to the initial direction of CoM movement and thereby serving as a counterweight to change the position of the CoM away from its initial direction (Allum *et al.*, 2002). Another restoring effect of the arm movements might be the accompanying centrifugal forces. Which of these components contributes the most to balance corrections and how large their magnitude is compared to other balance correcting responses is not known.

In the last decade Induced Accelerations Analysis (IAA) has shown to be a promising technique for analyzing the causes of movement in pathological gait (Neptune *et al.*, 2001). In IAA the effect of a joint torque and/or muscle force on the acceleration of the different joint angles, body segments and of the CoM is calculated. A key element of IAA is that it not only assesses the effect of a joint torque on its adjoining segments but on all segments of the model. By using IAA new insights have been gained regarding the contribution of plantar flexors (Neptune *et al.*, 2001) and upper leg muscles (Zajac *et al.*, 2003) to body support and forward progression. In particular, IAA has provided insight on the cause of decreased knee flexion during the swing phase in stiff knee gait of cerebral palsy patients (Goldberg *et al.*, 2003; Anderson *et al.*, 2004; Goldberg *et al.*, 2004), and adaptation in the knee and hip to compensate for the decreased ankle function in a stroke patient (Higginson *et al.*, 2006). Although this method has yet to be applied to studies of balance control, it offers the potential to improve our understanding of the factors that contribute most to restoring equilibrium.

Therefore, the goal of this study is to investigate the use of IAA to reveal the causes of body movement. A new method to calculate the induced accelerations has been developed and applied to calculate the contributions of the different joints, platform and arm movements to restore balance in response to platform perturbations. The method was based on analytically derived expressions, in contrast to the numerical approaches used in the gait studies described before. When applying IAA to study the balance responses, the induced horizontal accelerations of the CoM of each component can be used to derive the effect of the concerned component in stabilization of the CoM.

3.3 materials and methods

3.3.1 subjects

The subject was a 24 year old woman, who was 1.65 m tall and weighed 58 kg. She did not have any neuromuscular disorders. Before the experiment she signed an informed consent. The experiment was approved by the local ethics committee and was in accordance with the Declaration of Helsinki.

3.3.2 experimental apparatus and recordings

The subject stood on two force plates (Bertec, USA) fixed to the top of a moving belt (Model 5288, Boy Transport Material, Denmark) that was driven by a 5.5-kW motor (KMER 132.Mx6, VEM, Germany) and controlled by an automation driver (VLT 5000, Danfoss, Denmark). With this set-up, the force plate could be rapidly translated up to a maximum distance of 2 m in either the forward or backward direction. The subject assumed a comfortable stance, with an inter-heel distance of 17 cm, let their arms hanging freely at their sides and had their eyes open and fixed on a target approximately 3 m in front of them. Ear phones were worn to prevent any audio feedback from the motor or actions of the operator prior to the onset of the perturbation.

Platform acceleration was recorded with an accelerometer (Kistler, K-Beam 8302A10, USA) fixed to the front of the force plate. The accelerometer signal was low-pass filtered at 500 Hz (NL 125, Digitimer, UK), and sampled at 1 kHz using a CED micro 1401 and Spike2 data collection software (Cambridge Electronic Design, UK). The signal was subsequently digitally filtered using a 100 Hz low-pass recursive Butterworth filter. The onset of platform movement was determined as the first inflection of the horizontal acceleration for each individual trial.

Anterior–posterior moments and forces from the force plates were sampled at 1 kHz using the same collection set-up. Kinematic data were recorded using an eight-camera infrared motion analysis system (ProReflex, Qualisys, Sweden). Reflective markers were attached to the skin or tight fitting lycra shorts, over the following anatomical landmarks: bilateral heel, big toe, lateral malleolus, knee and greater trochanter, anterior superior iliac spine and the sacrum. Furthermore, markers were attached bilaterally to the shoulder, upper arm, elbow and wrist. Recording of position data was triggered 1 s prior to platform perturbation and measured for 10 s. Position data were sampled at 100 Hz, converted off-line to 3D coordinates, and digitally filtered using a 10 Hz low-pass recursive Butterworth filter.

3.3.3 experimental protocol

Postural responses were recorded during a set of 40 trials that consisted of an equal number of forward and backward perturbations as well as an equal number of short and long perturbations. All the different perturbations were randomly ordered. In this paper we will present the results of the short forward and backward perturbation, so a total of 20 trials were analyzed. The perturbations consisted of a surface translation with an initial acceleration (duration 300 ms; peak 1.3 m.s-2) followed after a 100 ms delay, by a deceleration pulse (duration 200 ms; peak 1.7 m.s-2). The maximal velocity, reached during the 100 ms delay, was 0.25 m.s-1 and the total displacement was 0.08 m.

3.3.4 data analysis

pre-processing of kinematic and kinetic data

The measured ground reaction forces and torques were compensated for the inertia of the top plate of the force plate (procedures according to(Preuss & Fung, 2004)) and resampled to a frequency of 100 Hz. Subsequently, the corrected torques and forces were low pass filtered with second order Butterworth filter with a cut off frequency of 10 Hz. From the filtered forces the Centre of Pressure (CoP) and the resultant ground reaction force of both force plates together was calculated. The joint angles and segment centre of masses were calculated from the measured marker positions in a bottom up approach, starting with the foot, subsequently the lower leg and so on. In calculating the segment positions and orientations, the anatomical properties were kept constant and the rotations were calculated based on the predefined rotation axes for each joint.

The subject's body was modeled with an inverse and forward model (for a schematic overview of the models and their interconnections see Figure 3.1). The inverse model was used to calculate the joint torques, which served as input for the forward model to calculate the induced accelerations. The anthropometric dimensions of both models were exactly the same and were preserved during every single time step of the analysis. The anthropometric data (mass, the position of the center of mass in the local frame and the moment of inertia tensor) were determined with the regression equations of Chandler *et al.* (Chandler *et al.*, 1975). These relations depend on the total body weight and the dimensions of the segments. The latter and the orientation of the local coordinate frame were determined from the marker positions in a posture in which the subject was standing straight up, according to the method described by Brand *et al.* (Brand *et al.*, 1982). The inverse model was made of a foot, shank, thigh, pelvis and trunk segment and left and right lower and upper arm (Figure 3.1). The mass and inertia of the hands and head were fused with the lower arms and trunk, respectively. The forward model was made of the same segments apart of the foot, which was kept out of this model for reasons explained later. As the movements of the legs and trunk occurred almost only in the

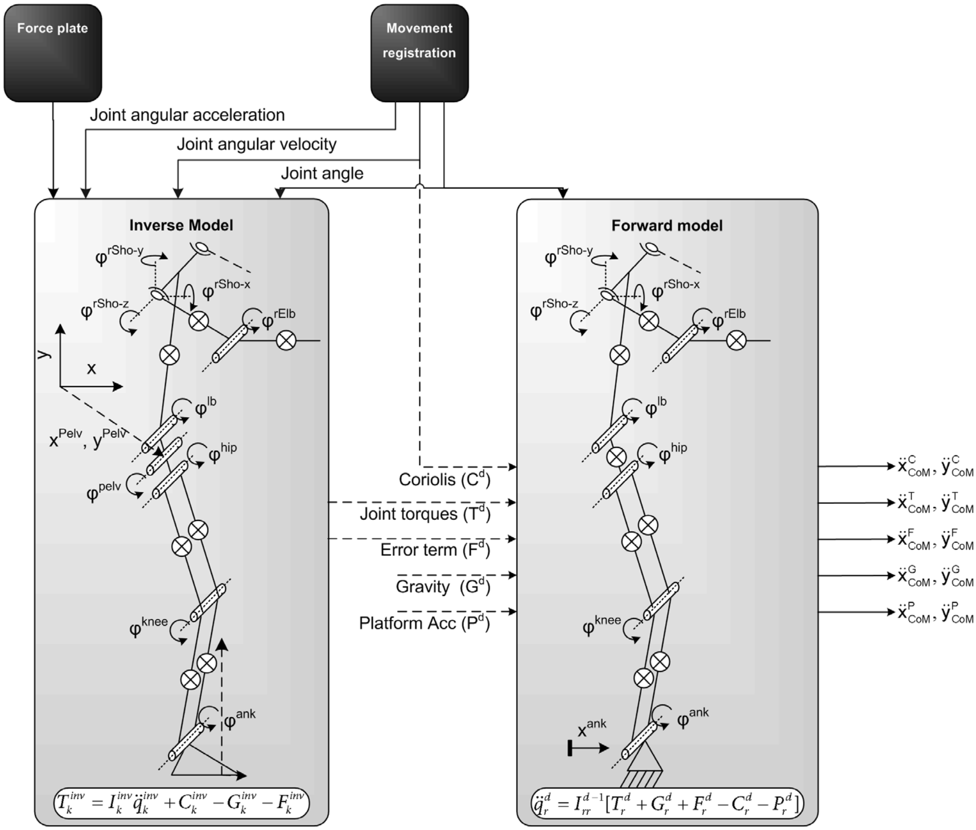

Figure 3.1. Inverse model (left) and forward model (right) with their degrees of freedom. The main difference between the degrees of freedom is the root of the model which is the position and orientation of the pelvis in the inverse model and the horizontal ankle position in the forward model. The measured movements and ground reaction forces serve as input in the inverse model to calculate the joint torques and the error term, which in turn serve as input together with gravity forces, platform acceleration and centrifugal forces to the forward model. Also the joint angles are input to the forward model as these are necessary to calculate the reduced mass matrix (I^u_{rr}). The left arm is not depicted but is part of both models and the definition of the degrees of freedom is similar to the definition of the right arm.

sagittal plane, the joints of these segments were modeled as hinge joints with a mediolateral axis. The shoulder joints were modeled as ball and socket joints because responses to the perturbations involved considerable arm movements out of the sagittal plane. The elbow joints were modeled as hinge joints. In short, the trunk, pelvis and legs are modeled in 2D while the arms are modeled in 3D.

The masses and inertias of the two feet, shank and thigh segments of both models were combined into single segments to prevent problems during the calculations with the forward model. The main difference between the inverse and forward model was the root of the model and the accompanying difference in the definition of some of the degrees of freedom of the model. The root of the model is the point in the human body with respect to which movements of all other segments are defined. The root of the model differed between models to make it possible to calculate an error term with the inverse model and to automatically get the interaction forces in the ankle for the forward model. These notions are explained in more detail in the paragraphs about the calculation of joint torques and about calculation of the induced accelerations respectively.

calculation of joint torques

The inverse model had a total of 15 degrees of freedom (Figure 3.1). The first 3 degrees of freedom were used to define the root of the model which was the position and orientation of the pelvis segment to the global frame. From this root the other segments branched off in two directions, towards the hands (upper branch) and towards the feet (lower branch). Three degrees of freedom, being the orientation of the upper leg, lower leg and foot with respect to the proximal segment defined the orientation of the lower branch. The remaining degrees of freedom described the movement in the upper branch and were the orientation of the trunk, lower arm and upper arm with respect to the proximal segment. We did not assume a left right symmetry between the arm movements because there was a considerable difference between the movements. Allowing unequal movements might result in an imbalance between the reaction forces in the shoulder along the sagittal (z) axis. As the segments below the shoulder are only allowed to move in the sagittal plane, this force will be transmitted to the ankle via the joint constraint forces.

The joint torques were calculated from the distal ends towards the root. The forces and accelerations in the one branch were not part of the calculations in the other branch. This implies that the joint torques in the lower branch were calculated on bases of the measured ground reaction forces and accelerations of the lower leg segments, while the joint torques in the upper branch were calculated solely on bases of the accelerations of the upper body and arms. Due to errors in these calculations (due to modeling assumptions about e.g. the position of the joint centers of rotation and the segment centers of mass) it could be that the sum of the resulting torques and forces of the lower and upper branch on the pelvis did not match with the measured accelerations. Also for multi-segment bodies, Newton's second law should apply which states that the sum of the forces is equal to the mass-acceleration product. The error term that had to be applied to the root to fulfill the equations of motions was a direct measure of the accuracy of the calculations.

The derivation of the inverse dynamical equations can be found in the appendix. The final equation used to calculate the forces and torques in the degrees of freedom has the measured movements and forces as input and the torques and forces in the degrees of freedom (T_k^{inv}) as output (see Figure 3.1) .

$$T_k^{inv} = I_k^{inv}(q_k^{inv})\ddot{q}_k^{inv} + C_k^{inv}(q_k^{inv}, \dot{q}_k^{inv}) - G_k^{inv}(q_k^{inv}) - F_k^{inv}(q_k^{inv}) \qquad (1)$$

where F_k^{inv} G_k^{inv} and C_k^{inv} are vectors with the external forces and torques, gravitational forces , centrifugal forces expressed in the degrees of freedom of the inverse model. I_k^{inv} is the reduced mass matrix, q_k^{inv} , $\dot{q}_k^{inv}$, $\ddot{q}_k^{inv}$ are the vectors with the positions, velocities and accelerations in the degrees of freedom and k indicates the number of degrees of freedom. The ground reaction forces and torques in the sagittal plane are part of the external force vector. T_k^{inv} contains the calculated torques in the joints as well as the forces and torques in the pelvis. The latter can be regarded as the error term.

calculation of induced accelerations

The forward model had a total of 12 degrees of freedom. This is 3 less than the inverse model, because the degrees of freedom necessary for the calculation of the error term were left out of the forward model, as the error term were inputs in the forward model (F_r^d see (2)). The root of the model was the movement of the ankle with respect to the global frame. The ankle was considered to be fixed to the platform. As a consequence applying a torque on the model automatically led to the corresponding joint reaction force in the ankle as this force was necessary to enforce the kinematic constraint. Consecutive application of the different torques in isolation led to a decomposition of the ankle joint reaction force into its different contributors. As the movements in the ankle were predefined and could not be changed by torques or forces generated in the human body, this degree of freedom was considered to be kinematic. The remaining degrees of freedom defined the orientation of the lower leg, upper leg, trunk, lower arm and upper arm with respect to the proximal segment and could be considered to be the dynamic degrees of freedom. For a more detailed explanation about the kinematic and dynamic degrees of freedom and to see the derivation of the equations of the forward model we refer to the appendix.

The accelerations in the dynamic degrees of freedom of the human body can be calculated with:

$$\ddot{q}_r^d = I_{rr}^{d\,-1}[T_r^d + G_r^d(q_r^d) + F_r^d - C_r^d(q_r^d, \dot{q}_r^d) - P_r^d] \tag{2}$$

where I_{rr}^d is the reduced mass matrix, T_r^d are the joint torques in the dynamic degrees of freedom (denoted with the d), G_r^d are the gravitational forces, F_r^d are the external forces, C_r^d are the coriolis and centrifugal forces and P_r^d are the equivalent torques in the degrees of freedom as a result of the platform acceleration. q_r^d, $\dot{q}_r^d$ and $\ddot{q}_r^d$, are the vector with positions, velocities and accelerations in the dynamic degrees of freedom, and r indicates the number of dynamic degrees of freedom (11). The terms of (2) have strong resemblance to the terms of (1), though the terms are now expressed in the dynamic degrees of freedom of the forward model and the external force term now consist of the error term and there is one extra term P_r^d.

In the IAA we want to calculate the effect of each of the joint torques, coriolis and centrifugal forces, gravitational forces and platform acceleration on the accelerations of the dynamic degrees of freedom of the forward model. For example the induced acceleration due to the joint torques are:

$$\ddot{q}_r^{d,T} = I_{rr}^{d\,-1} T_r^d \tag{3}$$

The accelerations in the dynamic degrees of freedom can be thought of as a summation of the induced accelerations of the separate terms, as indicated with a superscript in the following equation:

$$\ddot{q}_r^d = \ddot{q}_r^{d,T} + \ddot{q}_r^{d,G} + \ddot{q}_r^{d,F} + \ddot{q}_r^{d,C} + \ddot{q}_r^{d,P} \tag{4}$$

When the accelerations of the dynamic degrees of freedom are known, the rotational and translational accelerations of the segments with respect to the ankle can be calculated by pre-multiplying each term with the jacobian, $R_{i,r}^d$ (see also appendix (20)). For the joint torques:

$$\ddot{s}_i^{d,T} = R_{i,r}^d \ddot{q}_r^{d,T} \tag{5}$$

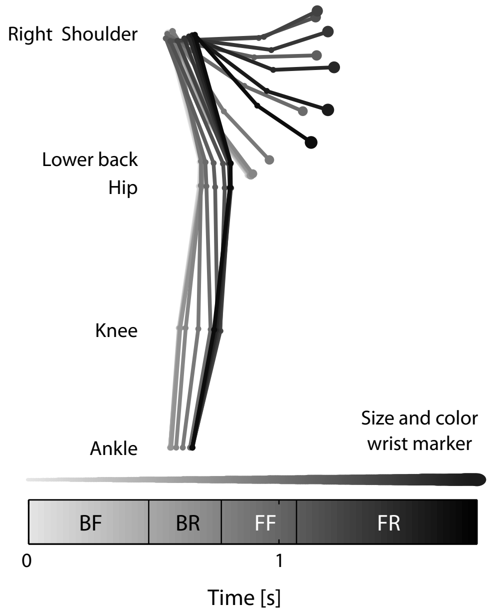

Figure 3.2. Displacements of the different joints in response to the forward perturbation for a single trial. Color coding and the size of the wrist marker reflect the time course of events. The different phases (as defined in Figure 3) of the response are indicated in the horizontal color bar. In response to the platform translation during the BF and start of BR phase, all segments initially showed a backward rotation. There is a clear distal to proximal progression in the onset of the rotation. After the initial counter clock wise rotation the ankle showed a clockwise rotation to counteract the translation. The arms swung forward during the BF and BR phase and returned during the FF and FR phase.

where, $\ddot{s}_i^{d,T}$ are the induced accelerations (translational and rotational) of the joint torques on the different segments. From the induced accelerations of the separate segments, the induced acceleration of the CoM can be calculated, by taking the weighted averages of the segment masses. The horizontal induced acceleration of the CoM will be indicated with $\ddot{x}_{CoM}^{T}$ and the vertical acceleration with $\ddot{y}_{CoM}^{T}$. For the ease of notification the d is left out of abbreviation and T can be replaced with the symbols indicating the other terms (i.e. $\ddot{x}_{CoM}^{C}$, $\ddot{x}_{CoM}^{P}$).

The accelerations in the degrees of freedom can be thought of as the summation of the separate terms (see (4)). Each of these terms can also be thought of as the sum of its separate sub terms, i.e. the induced acceleration of the joint torques is the sum of the induced accelerations of the ankle torque, knee torque, hip torque and so on. In order to maintain balance the induced accelerations of the joint torques have to counteract the disturbing induced accelerations from the ankle/platform acceleration and gravity. The effect of the platform acceleration and gravitational forces was assessed by applying each of these terms to the model. The induced accelerations of the arm movements on the CoM were calculated by applying the centrifugal forces as a result of the arm movements ($\ddot{x}_{CoM}^{C,arm}$, $\ddot{y}_{CoM}^{C,arm}$), the gravitational forces of both arms ($\ddot{x}_{CoM}^{G,arm}$, $\ddot{y}_{CoM}^{G,arm}$) and the generated torques in the elbow and shoulder of both arms ($\ddot{x}_{CoM}^{G,arm}$, $\ddot{y}_{CoM}^{G,arm}$) to the model. The induced accelerations were calculated for each trial. The average induced accelerations were calculated by point-by-point averaging of the successful trials.

3.4 **results**

The results of the forward platform perturbations will be presented and interpreted in detail. In the last paragraph we will briefly show the results of the backward platform perturbations. Forward platform perturbation

3.4.1 **stabilization of the Centre of Mass**

In response to the forward platform translation, the whole body initially "fell" backward, followed after about 0.5 s by a forward flexion of the arms (Figure 3.2). In the course of the response, the body and arms moved back to their original location. The movements of the body were reflected in the CoM movements (Figure 3.3b-d). As shown in Figure 3.3a and b, there were four distinct phases of CoM movement. These phases were defined by the sign of the CoM with respect to its initial value and the direction of the movement. Initially the CoM moved backward (backward fall "BF") followed by a return to the position prior to the perturbation (backward return "BR"). The movement did not halt at the initial position but passed this position which resulted in a forward fall "FF". Eventually, this forward fall was slowed down and the CoM moved back to its initial position (forward return "FR").

With IAA, the CoM acceleration is decomposed in its different contributors. Each of contributors stabilizes or destabilizes the CoM. The required direction to stabilize the CoM depends on the phase. In general, the induced acceleration of a term should be opposite to the CoM movement during either of the fall phases to counteract the movement and restabilize the CoM. During the return phases, the induced acceleration should initially be directed towards the equilibrium position to generate the required returning velocity, however at the end of the return phase the induced acceleration should reverse sign to slow down the returning movement and prevent an overshoot.

Judging from the similarity between the pattern and magnitude of the induced accelerations of the platform acceleration ($\ddot{x}_{CoM}^{P}$) and the accelerations of the CoM, the platform acceleration was clearly one of the main contributors to the horizontal acceleration of the CoM (Figure 3.4a). The similarity between both indicates that the platform acceleration not only destabilized the body during the BF phase but also largely contributed to the restabilization of the CoM during the second part of the BF phase and the BR phase. At the end of the BF phase and during the BR phase, the platform was decelerating which resulted in a $\ddot{x}_{CoM}^{P}$ opposite to the CoM movement. One could even argue that the deceleration of the platform contributed so much to the returning velocity that the human body was not able to stop at the initial position but made an overshoot. In this respect, the deceleration could be regarded as a second perturbation, which is in the opposite direction of the initial perturbation.

In order to counteract the disturbing effect of the platform acceleration the human body has to generate a response with an induced acceleration oppositely in sign of the platform term. The induced acceleration of the sum of generated joint torques ($\ddot{x}_{CoM}^{T}$) showed an increase of its baseline value for the forward perturbation, which would counteract the effect of the platform acceleration and cause a return of the CoM to its initial position. This increase lasted for the entire duration of the BF phase. In the remaining phases, the $\ddot{x}_{CoM}^{T}$ showed a decrease with respect to its baseline value. In the BR phase, this served to break the returning movement. However, at the end of the BR phase the CoM velocity was still close to its maximum (Figure 3.3). The subject moved with a great velocity through the equilibrium position and the subject fell

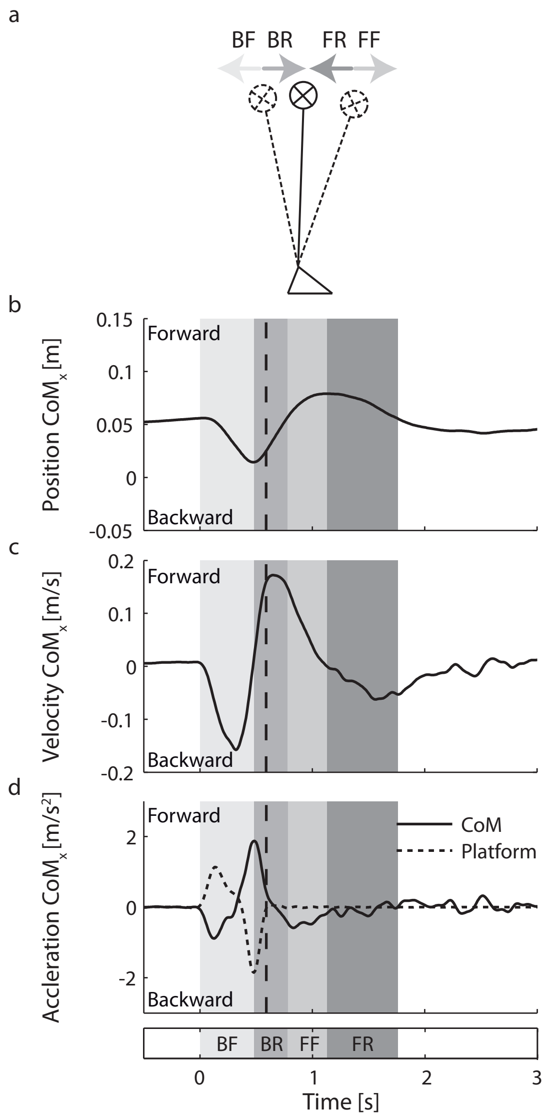

Figure 3.3. a) Definition of the different phases based on the horizontal CoM movements with respect to the ankle. BF indicates backward fall, BR backward return, FF forward fall and FR forward return. The position, velocity and acceleration of the horizontal Centre of Mass position are depicted in panel b, c and d respectively. In the plot of the CoM acceleration, also the platform acceleration is depicted. The onset of the perturbation is at time 0 s and the offset of the perturbation is indicated with the dashed vertical line. The different shadings indicate the different phases which are indicated in the horizontal bar on the bottom of the figure.

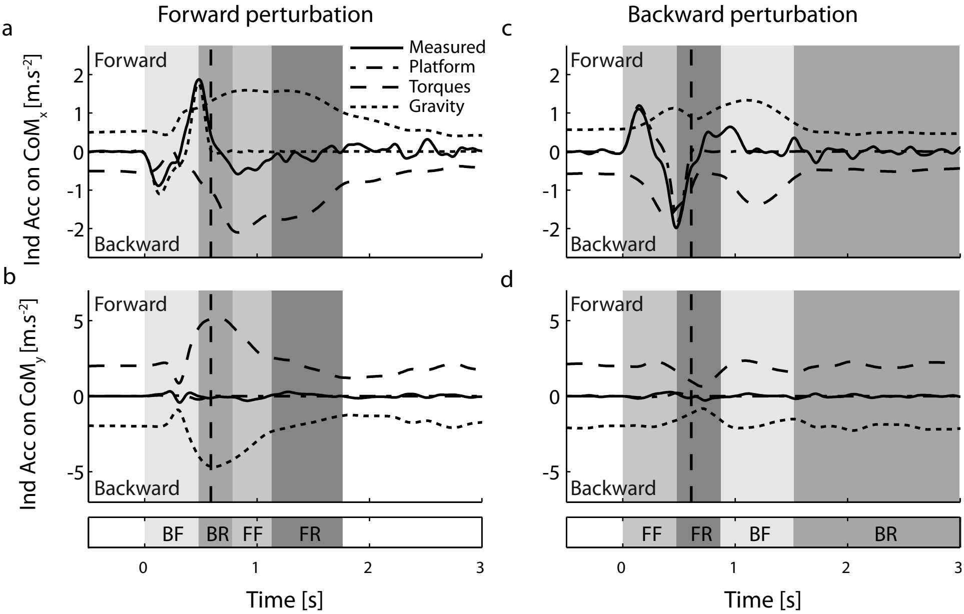

Figure 3.4. The horizontal (upper panels) and vertical (lower panels) induced accelerations on the CoM of the sum of the gravity forces ($\ddot{x}^G_{CoM}$, $\ddot{y}^G_{CoM}$), the sum of the joint torques ($\ddot{x}^T_{CoM}$, $\ddot{y}^T_{CoM}$) and the platform acceleration($\ddot{x}^P_{CoM}$, $\ddot{y}^P_{CoM}$) for the forward (left panels) and backward perturbation (right panels)

forward. The subject's body can be regarded as an underdamped inverted pendulum, resulting in a continued sway with overshoot after a perturbation occurred. The induced accelerations of the torques in the FF and FR phase first slowed down this movement and then set in the return to the initial position.

The joint torques not only had to counteract the disturbing platform translation but also the gravitational forces. The changes in induced accelerations of the gravitational forces are the consequence of the changes in the body orientation. After the onset of the perturbation $\ddot{x}^G_{CoM}$ first showed a very small drop and subsequently a large increase that lasted for the BR, FF and FR phases (Figure 3.4b). The contribution of the induced accelerations of the gravitational forces became especially clear after the platform translation ended. The sustained high positive induced accelerations contributed largely to the high positive CoM velocity, necessitating the joint torques to cause the induced accelerations in the opposite direction.

In the previous paragraphs, we considered the combined actions of all the joint torques. This induced acceleration is the sum of the induced accelerations of the separate joint torques. The subject initially generated a dorsiflexion, knee flexion, hip extension and lower back extension torque (Figure 3.5). After the perturbation ended, the joint torques often crossed their baseline levels, before returning to these levels. The pattern of the induced accelerations of each of the joint torques was similar to the pattern of the observed joint torques. However the "gain" (the reduced mass matrix) between the joint torques and induced accelerations differed between the joint torques as it was depended on the body configuration. $\ddot{x}^{T_{ank}}_{CoM}$ increased during the BF and BR phases and showed a decrease with respect to its baseline value during the FF and FR

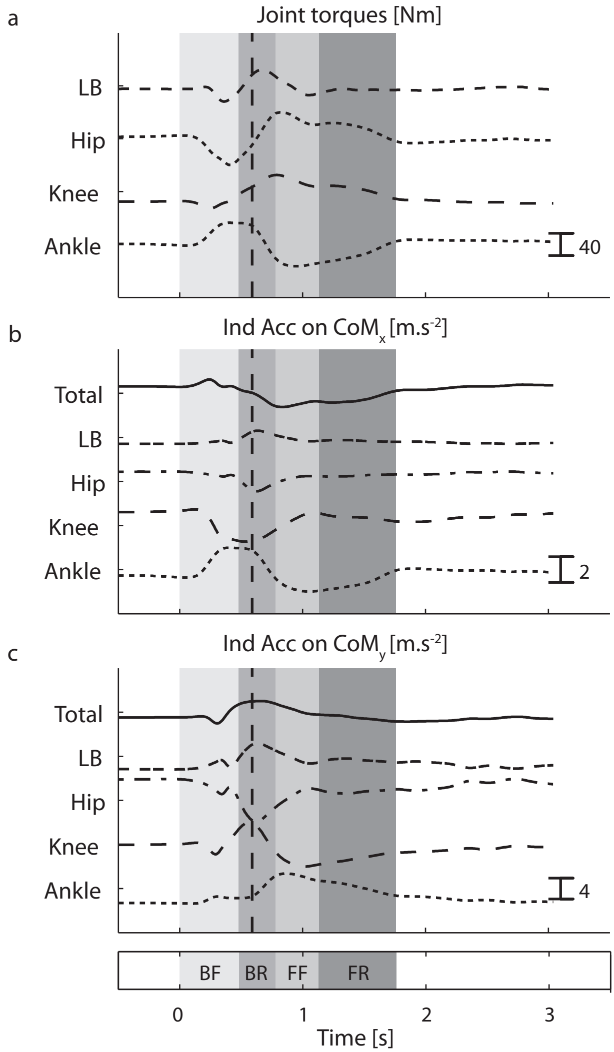

Figure 3.5. Joint torques (a) in the ankle, knee, hip and lower back and the horizontal induced accelerations, $\ddot{x}_{CoM}^{T_i}$, (b) and vertical induced accelerations, $\ddot{y}_{CoM}^{T_i}$, (c) on the CoM of these joint torques. Positive joint torques indicate dorsal flexion, knee extension, hip flexion and trunk bending. As a reference the induced accelerations in horizontal and vertical direction of the sum of the joint torque

phases. So in both phases it contributed largely to the stabilization of the CoM. In contrast, $\ddot{x}_{CoM}^{T_{knee}}$ showed the opposite pattern and therefore contributed negatively to the horizontal stabilization. One could question why the CNS would generate such a large joint torque that destabilizes the body. The answer to this question will follow when we present the results for the body weight support. The induced acceleration of the hip joint and lower back were relatively small compared to the ankle and knee and were opposite to each other. Some joints showed opposing induced accelerations, as was the case for the knee and ankle. In general, the induced acceleration of all the joint torques was dominated by the contribution of the ankle torque.

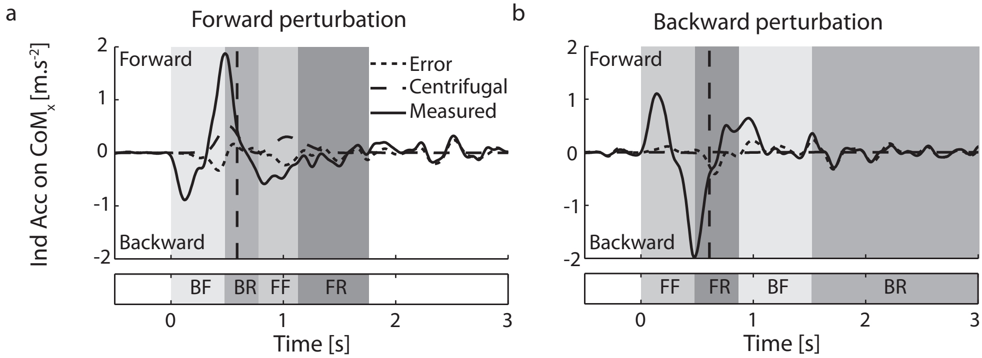

Figure 3.6. The induced accelerations on the horizontal CoM of the error term ($\ddot{x}_{CoM}^{F}$) and the centrifugal forces ($\ddot{x}_{CoM}^{C_i}$) for the forward (a) and backward perturbation (b). As a reference the measured accelerations of the CoM with respect to the ankles is depicted.

The error term is the result of the mismatch between the measured ground reaction forces and accelerations of the human body. The magnitude of the induced accelerations of the error term gives an indication about the reliability of the used model and the reliability and accuracy of the measurements. Compared to the measured acceleration the induced acceleration of the error term is generally small, however compared to the other terms its contribution was substantial (Figure 3.6a). The pattern is rather random and contains high frequency components so a detailed description of its contribution during the different phases is not relevant.

The induced accelerations of the centrifugal forces showed two small peaks in the positive direction during the transition between the BF and BR phase and during the FF phase (Figure 3.6a). The first peak was in the direction of the returning movement of the CoM and contributed with the joint torques to the reversal of the CoM velocity. Although the second peak was also positive it did not contribute to the return of the CoM but to the fall as it occurred during the forward fall.

The centrifugal forces showed a clear contribution to the stabilization of the CoM for forward perturbations. A possible explanation for the high contribution could be the swinging of the arms. Indeed, the arms swung forward in response to the forward perturbation (Figure 3.2). Apart from the large centrifugal forces, raising the arms could also affect the accelerations of the CoM through the generated shoulder and elbow torques and through the changes of the relative position of the gravity forces. The contributions of the arm joint torques and especially gravity are relatively small compared the contribution of the centrifugal forces of the arms (Figure 3.7). The two peaks in the induced accelerations of the total centrifugal forces in response to the forward perturbation were largely caused by raising and lowering the arms judging from the similarity between the timing and pattern of the peaks. As already discussed in more detail for the total centrifugal forces, the first peak contributed positively and the second peak negatively to the stabilization of the CoM. An indication about the contribution of each of the peaks can be obtained by integrating the areas below the curves. The area below the first peak of $\ddot{x}_{CoM}^{C,arm}$ was slightly larger than below the second peak, meaning that the centrifugal forces contributed more to stabilization during BF and BR than to destabilization during FF and FR. In contrast, on bases of the areas below the curves of $\ddot{x}_{CoM}^{T,arm}$ and $\ddot{x}_{CoM}^{G,arm}$, the arm torques and gravity contributed more to destabilization during the FF and FR phase than to stabilization during the BF and BR phase.

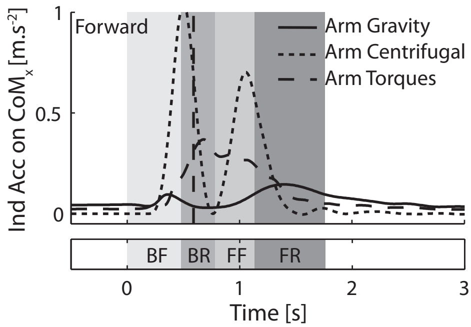

Figure 3.7. The induced accelerations on the horizontal CoM of the sum of gravity forces acting on the arm,($\ddot{x}_{CoM}^{G,arm}$), the sum of the generated torques in the shoulders and elbows ($\ddot{x}_{CoM}^{T,arm}$) and the centrifugal forces resulting from the arm movements ($\ddot{x}_{CoM}^{C,arm}$) in response to the forward perturbation.

3.4.2 weight support

In response to the perturbation the vertical position of the CoM also needs to be stabilized to provide body weight support. Although the accelerations of the CoM in vertical direction were relatively small Figure 3.4b), gravity and joint torques contributed significantly to the induced accelerations.

The induced accelerations of the gravitational forces ($\ddot{y}_{CoM}^{G}$) were continuously negative; in the middle of the BF phase it increased towards zero and subsequently it showed a large drop during the BR and FF phases. The changes in $\ddot{y}_{CoM}^{G}$ coincided with similar but opposite change in $\ddot{y}_{CoM}^{T}$. So, the joint torques were largely responsible for supporting the body weight. The joint torques at the knee and lower back were the main contributors to the increase of $\ddot{y}_{CoM}^{T}$ during the BF and BR phase (lower panel Figure 3.5c). So the joint torque in the knee was generated for the stabilization of the CoM in vertical and not in horizontal direction.

3.4.3 backward platform perturbation

In response to the backward perturbations, the body moves in a similar but opposite pattern (Figure 3.4c). However, after the overshoot the body moved back to its initial position very slowly resulting in a BR phase which extended after 3 seconds (see phase definition below Figure 3.4d). The results of induced acceleration analysis generally confirmed the results of the forward perturbations. The platform acceleration and deceleration dominated the accelerations of the CoM. The generated joint torques counteracted the destabilizing effect of gravity and the platform acceleration. Although not depicted, the ankle torque was again the main contributor to stabilization, while the knee joint torque contributed to destabilization and the hip joint and lower back were again relatively small. We also observed a remarkable difference. In contrast to forward perturbations, the centrifugal forces contributed little to either the stabilization or the destabilization during backward perturbations. This could be explained by the absence of significant arm movements in response to the backward perturbation.

3.5 discussion

The goal of this paper was to reveal the cause of movements in response to balance perturbations by analytically deriving the equations to calculate the induced accelerations. The results showed that the platform translation is the main contributor to the destabilization but also the initial restabilization of the CoM. Furthermore we showed that swinging the arms can contribute positively to the stabilization of the CoM through the centrifugal forces.

3.5.1 assumptions in the calculation of the induced accelerations

In calculating the induced accelerations several assumptions were made which merit further discussion. We assumed that the feet were fixed to the platform, so the accelerations of the ankle were equal to those of the platform. Although we studied foot in place reaction, it was possible that the front of the feet lifted up in response to the perturbation and consequently the ankle moved a bit in horizontal and vertical direction. The accompanying accelerations were not taken into account in the calculation of the accelerations of the different segments, which may have contributed to a mismatch between the measured forces and the calculated accelerations at the pelvis segment and consequently, added to the error term in horizontal and vertical direction. However, the induced accelerations of the total error term were relatively small, so the assumption that the feet were fixed to the platform seems to be justified.

In the models, we chose to combine both legs in "one leg". This was done because the movements in the joints of the right and left leg were very similar. When the joint angles were calculated for the separate legs the root mean square values of the difference between the joint angles of the ankles, knees and hips were 1.3°, 2.5° and 1.8° respectively. Both legs were also combined in one leg to avoid problems with the closed chain formed by the two legs, pelvis and the floor. Modeling of the closed chain with a kinematic constraint on both ankles and similar movements on both sides can result in a singular or close to singular reduced mass matrix. This would have affected the accuracy of the outcome of the forward model. Furthermore, the use of the "one leg" model instead of a model with a closed chain did not affect the results. Consider a model with two separate legs and only sagittal movements. Both ankles are fixed to the platform so the closed chain only has 2 degrees of freedom, e.g. the right ankle and knee. Because of this configuration, movements in the right ankle will always be accompanied by equal movements in the left ankle. So applying a joint torque to this model will always be accompanied by similar induced accelerations in the corresponding joints of both legs. Consequently, the induced accelerations of the leg joints in the "one leg" model would be the same as the sum of the induced accelerations of the left and right leg joints of a model with the closed chain incorporated.

3.5.2 design of perturbation signals

In our experiment, the platform acceleration is the main contributor to the acceleration of the CoM in the horizontal direction during the course of the platform movement. It destabilizes the body with the initial acceleration but also helps to restabilize the body with its deceleration. The deceleration might even perturb the body in the direction opposite to the initial perturbation. These results underline the importance of the acceleration profile of the perturbation. This was also indicated in earlier studies (McIlroy & Maki, 1994; Runge *et al.*, 1999; Bothner & Jensen, 2001; Carpenter *et al.*, 2005). However, in these studies the importance of the acceleration

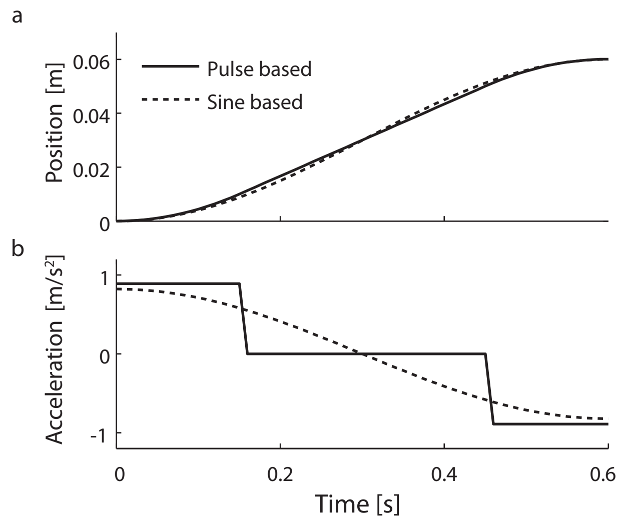

Figure 3.8. a) Two position and b) acceleration profile of two perturbation signals. The pulse based displacement signal is obtained by double integration of a pulse based acceleration profile, while the sine based signal is the increasing slope of a sine wave. The two signals have a very similar position profile, while their acceleration profile clearly differs

profile was shown by the coincidence of the start of acceleration and deceleration phases with EMG activity or reversals of joint and segment movements. In none of these studies, has the effect of the platform acceleration been quantified so unambiguously as in the current study.

Although, McIlroy and Maki (1994) stressed the need to consider the displacement, velocity, and acceleration characteristics of a support-surface perturbation independently, this has not become common practice (Chaudhuri & Aruin, 2000; Vearrier *et al.*, 2005). The necessity to describe the acceleration profile is clarified in Figure 3.8. The solid displacement signal is obtained by double integration of a pulse like acceleration profile, while the dashed signal is the increasing slope of a sine wave. The depicted perturbation signals have exactly the same displacement and approximately the same maximal velocity (Pulse: 0.134 $m.s^{-1}$, Sine: 0.157 $m.s^{-1}$) and maximal acceleration (Pulse: 0.89 $m.s^{-2}$, Sine: 0.82 $m.s^{-2}$), however their acceleration profiles clearly differ. In all probability, this will affect the postural response as the platform acceleration dominates the acceleration of the CoM. Brown and colleagues (Brown *et al.*, 2001) showed that the onset latency for the Gastrocnemius muscle, the time to maximal CoM excursion were smaller and the impulse of the ankle joint torque were larger in response to a backward pulse based perturbation compared to a sine based perturbation. The pulse based signal required a more intense response as the actual perturbation (the acceleration) was more concentrated at the start of the perturbation and the restabilizing effect of the deceleration was delayed. Unfortunately, Brown and colleagues only reported the results until the maximal CoM excursion was reached and not the effect of the acceleration profile on the return of the CoM to the equilibrium position. In our view, the time between the acceleration pulse and deceleration pulse should be maximized. Still, this should be combined with unpredictability in the onset and offset of the acceleration and deceleration pulse (McIlroy & Maki, 1994). So the subject is not only forced to generate a response to stabilize from the platform induced falling movement, but he can also not count on the platform deceleration for his return movement. The random use of a triphasic pattern, in which a second acceleration pulse can occur, would force the subject even more to generate an accurate genuine response to the first acceleration pulse(McIlroy & Maki, 1994).

3.5.3 applicability of the analytical approach to other situations

This was the first time induced accelerations have been used to determine the cause of movements in balance responses. Previous studies that applied IAA in gait (Neptune *et al.*, 2001; Zajac *et al.*, 2002; Goldberg *et al.*, 2003; Zajac *et al.*, 2003; Goldberg *et al.*, 2004; Higginson *et al.*, 2006), all used forward dynamical simulations in combination with optimization to determine the induced accelerations of the different muscles. Forward dynamical simulations were required in these studies to assess the activation and generated force of the muscles and for an appropriate modeling of the foot ground contact. In this study, we used the net joint torques instead of the muscle forces as input for the induced accelerations. Since, there was no roll off of the feet, we assumed that the feet were rigidly connected to the floor and that they only transferred the forces and torques around the ankle to the ground. With these assumptions, we could use an analytical approach to calculate the induced accelerations. This makes the use of induced acceleration applicable in a broader range of research and even clinical laboratories. Furthermore, with the analytical approach it was possible to assess the induced accelerations of the centrifugal forces, while this was not possible with the forward simulations. Though the contribution of the centrifugal forces is generally thought to be small, the rapid swinging of the arms in response to the forward translation caused centrifugal forces that clearly contributed to stabilisation of the CoM. In response to the backward translation, the contribution of the centrifugal forces was clearly smaller than observed in forward translation. These differences, coupled with unique joint torque contributions to induced accelerations observed for forward and backward perturbations clearly demonstrates the sensitivity of induced acceleration analysis to different conditions.

As mentioned before, previous studies using IAA concentrated on the contribution of the different muscle forces during gait. The application of the analytical approach to gait will encounter several difficulties. In gait, the ankle can not supposed to be fixed to the floor, as it is continuously moving with respect to the floor. Consequently, the foot has to be incorporated in the model and the foot-floor contact has to be modeled. Furthermore, during double support a closed chain arises in which the movements of both legs are not necessarily the same. Recently, Hof and Otten (2005) derived analytical expressions to calculate the induced accelerations during gait. The aforementioned difficulties made several assumptions necessary in their approach. However, the effects of these assumptions on analysis results are unclear. The major assumption was that modeling one leg, while representing the contralateral leg as independent force in the hip joint would be adequate. The induced acceleration of this "second leg"-force was separately calculated. In doing so, they neglected that part of this force is caused by the inertia and kinematical constraints of the second leg, as reaction to the torques/forces of which the induced accelerations are calculated. So part of the induced accelerations of the second leg should actually be considered as induced by the analyzed torques/forces. Furthermore, during the stance phase they assumed that at every single time step the acceleration of the CoP was equal to zero, which did not correspond to their measured data in which the CoP moves forward and shows acceleration at every single time step. Considering the CoP as a fixed joint neglects that every torque/force also induces an acceleration of the CoP location. This also changes the induced accelerations at other locations. Consequently, before analytically derived expressions of induced accelerations can be applied to gait, solutions have to be found to overcome the main difficulties and/or the effect of these necessary assumptions have to be assessed.

Induced acceleration analyses can also be applied in studying balance responses in different situations and in pathologies like stroke patients and prosthetic patients. For example when studying reach-to-grasp reactions, the model of the current study can be used to calculate the contribution of the redistribution of segment masses on the stabilization of the CoM as long as the hand has not yet grasped something. When the hand has grasped on to an object, the interaction forces should be measured and incorporated in the calculation of the arm joint torques and their induced accelerations by extending the inverse model and forward model, respectively. For pathologies, the contribution of the joint torques of the affected and non affected leg to stabilization of the CoM in response to platform translations can be assessed. As long as the movements in both legs are the same, only the inverse model has to be adapted to encompass both separate legs, so the joint torques in both legs can be calculated. The forward model does not need to be adapted as a model consisting of one leg with lumped masses and inertias is mathematically the same as a model with a closed chain consisting of two separate legs in exact the same posture. Both torques of a particular joint should be applied successively to the combined joint to calculate the separate induced accelerations. If the assumption of equal movements in both legs can not be held, the forward model and the inverse model get a lot more complex, as unequal movements in both legs can only be modeled by adding a degree of freedom outside the sagittal plane, like pelvic tilt.

3.6 appendix

3.6.1 derivation of inverse model

We used a modification of Kane's method to derive the equations of motion. In this method the principles of virtual power is used to rewrite the Newton-Euler equations. The second law of Newton states

$$\sum f_i - M_{ij}\ddot{s}_j = 0 \qquad (6)$$

where $\sum f_i$ is the sum of all forces and torques (at and around a given point respectively) on the 9 segment model, $\ddot{s}_j$ is a vector with the translational and rotational accelerations of all the separate segments and M_{ij}, is a i x j matrix with on the diagonal the mass and the inertia of the different segment.

Combining (6) with virtual velocities results in the virtual power of the system:

$$\partial\dot{s}_i\left[\sum f_i - M_{ij}\ddot{s}_j\right] = 0 \qquad (7)$$

where $\partial\dot{s}_i$ is the vector with the virtual velocities at the points of application of the forces and the virtual angular velocities in the direction of the torques.

Equation (7) will be rewritten in the generalized coordinates by expressing s and it derivatives in the generalized coordinates. The vector of generalized coordinates is called q_k^{inv} where k stands for the number of elements (in this case 15) and inv stands for inverse model

$$s_i = R_i^{inv}(q_k^{inv}) \tag{8}$$

where R_i^{inv} is a vector containing the positions and angles of s_i expressed in generalized coordinates. The velocities can be expressed as

$$\dot{s}_i = \frac{\partial R_i^{inv}}{\partial q_k^{inv}} \dot{q}_k^{inv} = R_{i,k}^{inv} \dot{q}_k^{inv} \tag{9}$$

where $R_{i,k}^{inv}$ is a two-dimensional matrix holding the partial differentials of R_i^{inv} with respect to the elements of a vector with index k. The comma indicates this partial differentiation.

The virtual velocities are

$$\partial \dot{s}_i = R_{i,k}^{inv} \partial \dot{q}_k^{inv} \tag{10}$$

The accelerations are given by

$$\begin{aligned} \ddot{s}_i &= R_{i,k}^{inv} \ddot{q}_k^{inv} + \frac{\partial R_{i,k}^{inv}}{\partial t} \dot{q}_k^{inv} \\ &= R_{i,k}^{inv} \ddot{q}_k^{inv} + \frac{\partial R_{i,k}^{inv}}{\partial q_l^{inv}} \frac{\partial q_l^{inv}}{\partial t} \dot{q}_k^{inv} \\ &= R_{i,k}^{inv} \ddot{q}_k^{inv} + R_{i,kl}^{inv} \dot{q}_k^{inv} \dot{q}_l^{inv} \end{aligned} \tag{11}$$

where $R_{i,kl}^{inv}$ is a 3 dimensional matrix with the partial differentials of $R_{i,k}^{inv}$ with respect to each of the generalized coordinates.

Substitution of (10) and (11) in (7) results in

$$R_{i,k}^{inv} \partial \dot{q}_k^{inv} \left[\sum f_i - M_{ij} \left[R_{i,k}^{inv} \ddot{q}_k^{inv} + R_{i,kl}^{inv} \dot{q}_k^{inv} \dot{q}_l^{inv} \right] \right] = 0 \tag{12}$$

This should hold for arbitrary virtual velocities $\partial \dot{q}_k$, consequently

$$R_{i,k}^{inv} \left[\sum f_i - M_{ij} \left[R_{i,k}^{inv} \ddot{q}_k^{inv} + R_{i,kl}^{inv} \dot{q}_k^{inv} \dot{q}_l^{inv} \right] \right] = 0 \tag{13}$$

The above equation consists of different terms, with:

$$R_{i,k}^{inv} M_{ij} R_{i,kl}^{inv} \dot{q}_k^{inv} \dot{q}_l^{inv} = C_k^{inv} \tag{14}$$

where C_k^{inv} are the coriolis and centrifugal forces expressed in the degrees of freedom.

$$R_{i,k}^{inv} M_{ij} R_{i,k}^{inv} \ddot{q}_k^{inv} = I_{kk}^{inv} \ddot{q}_k^{inv} \tag{15}$$

where I_{kk}^{inv} is the reduced mass matrix

$$R_{i,k}^{inv} \sum f_i = R_{i,k}^{inv} \left[f_i^{inv} + g_i + t_i \right] = F_k^{inv} + G_k^{inv} + T_k^{inv} \tag{16}$$

where f_i^{inv}, g_i and t_i are the one dimensional vectors with the external forces, the gravitational forces and the joint torques respectively. F_k^{inv}, G_k^{inv} and T_k^{inv} are the corresponding forces and torques expressed in the degrees of freedom of the inverse model.

Substitution of (14), (15) and (16) in (13) results in the general form of the movement equation

$$T_k^{inv} = I_k^{inv} \ddot{q}_k^{inv} + C_k^{inv} - G_k^{inv} - F_k^{inv} \tag{17}$$

With (17) the forces and torques in the degrees of freedom will be calculated, which is the purpose of the inverse model.

3.6.2 derivation of forward model

The purpose of the forward model is to calculate the accelerations from the known joint torques. The derivation of the forward model resembles the derivation of the equations used to calculate the joint torques, but deviates on certain important points.

The positions/angles and the velocity/angular velocity of the different segments can be expressed in the generalized coordinates of the forward model (q_p^{for}, where p stands for the number of degrees of freedom, 12) as

$$s_i = R_i^{for}(q_p^{for}) \quad (18)$$

$$\dot{s}_i = \frac{\partial R_i^{for}}{\partial q_p^{for}} \dot{q}_p^{for} = R_{i,p}^{for} \dot{q}_p^{for} \quad (19)$$

Next we need to make a distinction between the different degrees of freedom. In the induced acceleration analysis we want to assess the effect of joint torques on the accelerations of the degrees of freedom. However not all of the degrees of freedom can be influenced by the magnitude and direction of the different torques. The platform movement and consequently the ankle movement are prescribed and will remain the same irrespective of the applied torques. Therefore this degree of freedom is called a kinematic degree of freedom (indicated with the superscript kin, q_t^{kin}). The movements in the other degrees of freedom, which describe the movement of the body with respect to the platform movement, can be influenced and are regarded dynamic (indicated with the superscript d, q_r^d). The subscript t and r indicate the number of kinematic and dynamic degrees of freedom which are equal to 1 and 11 respectively.

$R_{i,p}^{for}$ matrix is also split up into a matrix containing the partial differentials of the kinematic degree of freedom and a matrix with the partial differential of the dynamic degrees of freedom, $R_{i,t}^{kin}$ and $R_{i,r}^{d}$ respectively. The accelerations can then be expressed in the kinematic (q_t^{kin}) and dynamic degrees of freedom (q_r^d):

$$\ddot{s}_i = R_{i,r}^d \ddot{q}_r^d + R_{i,rs}^d \dot{q}_r^d \dot{q}_s^d + R_{i,t}^{kin} \ddot{q}_t^{kin} \quad (20)$$

where $R_{i,rs}^d$ is a 3D matrix with the partial differentials of $R_{i,r}^d$ to each of the dynamic degrees of freedom. For the kinematic degree of freedom, the corresponding matrix consists of all zeros and is kept out of this and subsequent equations.

For the forward model (13) turns into:

$$R_{i,r}^d \left[\sum f_i - M_{ij} \left[R_{i,r}^d \ddot{q}_r^d + R_{i,rs}^d \dot{q}_r^d \dot{q}_s^d + R_{i,t}^{kin} \ddot{q}_t^{kin} \right] \right] = 0 \quad (21)$$

$\sum f_i$ is the sum of the gravitational forces, the net joint torques and the external forces. The gravitational forces are the same as in the inverse model. The net joint torques are the joint torques as calculated with the inverse model. The external forces are now the error term. The ground reaction forces do not have to be applied as in the forward model the ankle is constrained to a specified position. Applying torques or forces to the segments of the model will automatically lead to the joint forces in the ankle which keep the ankle in the specified position.

For the forward model (14), (15) and (16) turn into:

$$R^d_{i,r} M_{ij} R^d_{i,rs} \dot{q}_r \dot{q}_s = C^d_r \tag{22}$$

where C^d_r are the coriolis and centrifugal forces expressed in the dynamic degrees of freedom.

$$R^d_{i,r} M_{ij} R^d_{i,r} \ddot{q}^d_r = I^d_{rr} \ddot{q}^d_r \tag{23}$$

where I^d_{rr} is the reduced mass matrix

$$R^d_{i,r} \sum f_i = R^d_{i,r} \left[f^d_i + g_i + t_i \right] = F^d_r + G^d_r + T^d_r \tag{24}$$

where f^d_i, gi and ti are the one dimensional vectors with the external forces, the gravitational forces and the joint torques respectively and F^d_r, G^d_r and R^d_r are the external forces, gravitational forces and joint torques expressed in the dynamic degrees of freedom of the forward model.

In addition to the above described terms the forward model contains one extra term

$$P^d_r = R^d_{i,r} M_{ij} R^{kin}_{i,t} \ddot{q}^{kin}_t \tag{25}$$

where P^d_r expresses the contribution of the ankle acceleration, which is assumed to be equal to the platform acceleration, to the torques in dynamic degrees of freedom.

In the forward model we want to calculate the acceleration, therefore (22), (23), (24) and (26) are substituted in (21) and rewritten such that the accelerations are the only term on the left side. The final equation is:

$$\ddot{q}^d_r = I^{d\,-1}_{rr} [T^d_r + G^d_r + F^d_r - C^d_r - P^d_r] \tag{26}$$

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3.8 references

Allum JH, Carpenter MG, Honegger F, Adkin AL & Bloem BR. (2002). Age-dependent variations in the directional sensitivity of balance corrections and compensatory arm movements in man. *The Journal of Physiology* 542: 643-663

Anderson FC, Goldberg SR, Pandy MG & Delp SL. (2004). Contributions of muscle forces and toe-off kinematics to peak knee flexion during the swing phase of normal gait: an induced position analysis. *Journal of Biomechanics* 37: 731-737

Bogey RA, Perry J & Gitter AJ. (2005). An EMG-to-force processing approach for determining ankle muscle forces during normal human gait. *IEEE Transactions on Neural Systems and Rehabilitation Engineering* 13: 302-310

Bothner KE & Jensen JL. (2001). How do non-muscular torques contribute to the kinetics of postural recovery following a support surface translation? *Journal of Biomechanics* 34: 245-250

Brand RA, Crowninshield RD, Wittstock CE, Pedersen DR, Clark CR & van Krieken FM. (1982). A model of lower extremity muscular anatomy. *Journal of Biomechanical Engineering* 104: 304-310

Brown LA, Jensen JL, Korff T & Woollacott MH. (2001). The translating platform paradigm: perturbation displacement waveform alters the postural response. *Gait & Posture* 14: 256-263

Carpenter MG, Thorstensson A & Cresswell AG. (2005). Deceleration affects anticipatory and reactive components of triggered postural responses. *Experimental Brain Research* 167: 433-445

Chandler RF, Clauser CE, McConville JT, Reynolds HM & Y oung JW. (1975). Investigation of the inertial properties of the human body. Report DOT HS-80143:. National Technical Information Service, Springfield, Virginia.

Chaudhuri S & Aruin AS. (2000). The effect of shoe lifts on static and dynamic postural control in individuals with hemiparesis. *Archives of Physical Medicine and Rehabilitation* 81: 1498-1503

Goldberg SR, Anderson FC, Pandy MG & Delp SL. (2004). Muscles that influence knee flexion velocity in double support: implications for stiff-knee gait. *Journal of Biomechanics* 37: 1189-1196

Goldberg SR, Ounpuu S & Delp SL. (2003). The importance of swing-phase initial conditions in stiff-knee gait. *Journal of Biomechanics* 36: 1111-1116

Higginson JS, Zajac FE, Neptune RR, Kautz SA & Delp SL. (2006). Muscle contributions to support during gait in an individual with post-stroke hemiparesis. *Journal of Biomechanics* 39: 1769-1777

Hof AL & Otten E. (2005). Assessment of two-dimensional induced accelerations from measured kinematic and kinetic data. *Gait & Posture* 22: 182-188

McIlroy WE & Maki BE. (1994). The 'deceleration response' to transient perturbation of upright stance. *Neuroscience Letters* 175: 13-16

Neptune RR, Kautz SA & Zajac FE. (2001). Contributions of the individual ankle plantar flexors to support, forward progression and swing initiation during walking. *Journal of Biomechanics* 34: 1387-1398

Preuss R & Fung J. (2004). A simple method to estimate force plate inertial components in a moving surface. *Journal of Biomechanics* 37: 1177-1180

Runge CF, Shupert CL, Horak FB & Zajac FE. (1999). Ankle and hip postural strategies defined by joint torques. *Gait & Posture* 10: 161-170

Staudenmann D, Kingma I, Daffertshofer A, Stegeman DF & van Dieen JH. (2006). Improving EMG-based muscle force estimation by using a high-density EMG grid and principal component analysis. *IEEE Transactions on Biomedical Engineering* 53: 712-719

Vearrier LA, Langan J, Shumway-Cook A & Woollacott M. (2005). An intensive massed practice approach to retraining balance post-stroke. *Gait & Posture* 22: 154-163

Zajac FE, Neptune RR & Kautz SA. (2002). Biomechanics and muscle coordination of human walking. Part I: introduction to concepts, power transfer, dynamics and simulations. *Gait & Posture* 16: 215-232

Zajac FE, Neptune RR & Kautz SA. (2003). Biomechanics and muscle coordination of human walking: part II: lessons from dynamical simulations and clinical implications. *Gait & Posture* 17: 1-17

chapter 4

use of surface perturbations to increase the load on the paretic leg during balance training in stroke survivors

submitted

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4.1 abstract

The aim of this study was to assess whether training which addressed the postural control in the paretic leg and the ability to withstand perturbations could improve balance function and walking capacity in chronic stroke patients. We used a single-subject, non-concurrent, multiple baseline design with four chronic stroke survivors. During training, subjects counteracted perturbations consisting of surface translations moving them toward their paretic side or backward. With training, subjects were able to withstand larger perturbations. The larger perturbation magnitudes resulted in an increased demand on the paretic leg. However, in none of the subjects, was this accompanied by a clear increase in the contribution of the paretic leg to balance control. Only one subject demonstrated a clear improvement of the performance on clinical tests. In conclusion, the current balance training like other balance training regimes resulted in a specific effect on the trained ability and limited transfer to functional tasks. Because the subjects were at least one year post-stroke, possibly their compensatory use of the non-paretic leg influenced the training. In an (sub)acute patient population, the increased demand on paretic leg induced by the platform movement might be a sufficient trigger for restitution of function.

4.2 **introduction**

A common consequence of stroke is disturbed balance control arising from lesions in the central nervous system that affect motor and sensory aspects of movement and postural control. Different aspects of balance control are affected, including balance control during quiet stance, expressed by a decreased weight bearing on the paretic leg and forward lean (Nardone *et al.*, 2001; de Haart *et al.*, 2004) and balance control in anticipation of movement, expressed by decreased weight shifting abilities (Diener *et al.*, 1993; de Haart *et al.*, 2005). Consequently, gait and the performance of personal daily tasks are detrimentally affected and the risk of falling is increased (Sackley, 1991; Nyberg & Gustafson, 1997; Hyndman *et al.*, 2002; Jorgensen *et al.*, 2002). The presence of inadequate responses to external perturbations (Badke & Duncan, 1983; Ikai *et al.*, 2003) increase this risk further. Recently, Marigold and Eng (Marigold & Eng, 2006) demonstrated that a delayed onset of postural reflexes in response to a perturbation contributed to the occurrence of falls in chronic stroke patients.

In the last decades several training regimes have been developed and used to retrain balance in stroke patients (Aruin *et al.*, 2000). One of the most often used is biofeedback training (Shumway-Cook *et al.*, 1988; Winstein *et al.*, 1989; Walker *et al.*, 2000), in which patients receive visual and/or auditory feedback regarding the position of their Center of Pressure (CoP). The feedback can be specifically aimed to address steadiness (maintain the CoP within a narrow target or range), symmetry (maintain the CoP as close to the midline as possible) or dynamic stability (shift the CoP position in anteroposterior or mediolateral directions to selected targets). The different training regimes especially resulted in improvements of their respective training goals but did not relate to improvements of more functional aspects of balance that exceed the improvements achieved with more conventional therapy (Barclay-Goddard *et al.*, 2004).

Biofeedback training specifically focuses on balance control during quiet stance and in anticipation of movement. Although the capacity to withstand perturbations is a very important aspect of balance control and as such in preventing falls (Marigold & Eng, 2006), training strategies addressing this aspect of balance control are very rare. One reason for this might be that it is difficult to provide perturbations in a systematic, controlled and safe manner. This could be overcome by using a moving platform to apply the perturbations. Hitherto, only Hocherman and colleagues (Hocherman *et al.*, 1984) have made use of platform movements to train chronic stroke survivors. During the course of the training the subjects were able to withstand movements with greater amplitude, however any associated improvement on more functional scales of balance was unreported.

The main objective of our study was to design and evaluate a balance training for stroke survivors that specifically addressed their capacity to withstand surface perturbations by making use of platform movements. The evaluation was performed in a group of 4 chronic stroke survivors. Previous studies have demonstrated that post-stroke, people have difficulty withstanding perturbations inducing sway toward the paretic side (Holt *et al.*, 2000; Ikai *et al.*, 2003) and backward (Marigold & Eng, 2006). Since the majority of falls are directed towards the paretic side (Hyndman *et al.*, 2002), we concentrated on using perturbations during training which induced sway either toward the paretic side or toward the back. The experienced difficulties when swaying toward the paretic side can be explained from the fact that the sway is accompanied by an increase of load on the paretic leg, requiring that automatic responses of the paretic leg are necessary to counteract the sway. In case of backward sway, less movement of the CoP is allowed to control the sway, as the distance of the Center of Mass (CoM) to the posterior boundary of the base of support is generally smaller than to the anterior boundary. The limited CoP motion requires a faster response of the anterior leg muscles, which commonly show activation abnormalities post-stroke.

We aimed to improve the ability to withstand perturbations, by increasing the contributions of the paretic leg. The acceleration profile of the platform was designed such that a lateral perturbation "pushed" the body towards the paretic leg, eliciting the paretic leg to contribute to counteracting the disturbance. The accelerations in the anterior/posterior plane "pushed" the subject backward. Although, the backward perturbation did not specifically increase the load on the paretic leg, subjects were encouraged by the therapist to use their paretic leg as much as possible in withstanding the perturbation. By focusing on the postural contributions in the paretic leg, we hypothesized that the training could have a beneficial effect for the functional balance. This would be the case, when the improved control in counteracting external perturbations transferred to improved control to internal perturbations, which occur during self-initiated movements, such as walking, reaching and turning.

4.3 materials and methods

4.3.1 subjects

Four subjects with hemiparesis secondary to a single and first ever unilateral stroke in the territory of the anterior cerebral arteries participated (see Table 4.1). Subjects were at least one year post-stroke, had no musculoskeletal or neurological diseases in addition to stroke, were able to remain standing without help of support for at least 90 s and were able to adequately comprehend our instructions.

The experiment was approved by the local medical ethical committee and conformed to the principles of the Declaration of Helsinki. All subjects gave their written informed consent prior to the start of the experiment.

4.3.2 design

In this study we applied a non-concurrent multiple baseline design across subjects (Ottenbacher, 1986; Backman *et al.*, 1997; Backman & Harris, 1999). In a single subject design, the unit of study is an individual rather than a group and the individuals serve as their own control. Other important aspects of the single subject design are (1) that a baseline is carefully determined by several assessments and (2) that assessments are continued throughout the training period.

Table 4.1. Summary of subject characteristics.

Patient	Sex	Age	Months since stroke	Affected Hemisphere	Stroke type	AFO	DA
1	M	62	69	Right	ischemic	Yes	0.04
2	M	68	52	Left	ischemic	Yes	0.07
3	F	58	76	Right	hemorrhagic	No	0.07
4	M	51	23	Right	ischemic	Yes	0.06

AFO indicates the use of an Ankle Foot Orthosis. DA indicates the amplitude of the disturbance signal in meters.

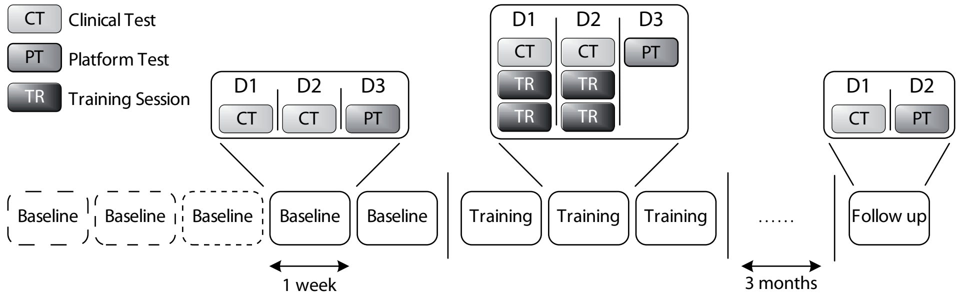

Figure 4.1. Schematic design of the study. The dashed boxes indicate that not all subjects performed this block of tests. The number of subjects performing the blocks decreased by one for each subsequent dashed block, which indicates the multiple baseline design of this study.

Apart of these characteristics, the non-concurrent multiple baseline distinguishes itself by the different lengths of the baseline period, which are randomly assigned to the different subjects. In this study the length of the baseline periods were 2, 3, 4 and 5 weeks.

Assessments were subdivided in platform tests and clinical tests (see Figure 4.1). The platform tests were performed once a week and the clinical tests twice a week. During the baseline period, all tests were performed on separate days. During the training period, the clinical tests preceded the training sessions. Between the clinical tests and training sessions subjects were given one hour to rest and prepare for the training. Follow-up testing was completed once approximately 3 months after the end of the training. In the following paragraphs the contents of the tests are described.

4.3.3 **platform test**

To monitor the changes the contribution of the paretic and non paretic leg to balance control, we assessed the Dynamic Balance Contribution (DBC) using perturbations of the platform. An extensive description of the used method can be found in Van Asseldonk *et al.* (2006). The following provides a brief description:

Subjects stood with their feet shoulder width apart on a 6 degrees of freedom motion platform (Caren, Motek, Amsterdam, The Netherlands). Balance responses were elicited by continuous platform movements, consisting of a multi-sine, in forward backward direction. The amplitude of platform movement was set to the maximal amplitude the subjects could comfortably withstand during the first test and was held constant throughout all subsequent platform tests. Body movements and ground reaction forces below each foot were recorded during 3 trials of 90 s.

Movements of the different body segments were measured by means of a three-dimensional passive registration system, consisting of six video cameras and a control unit (Vicon Oxford Metrics, Oxford, UK) at a sampling rate of 120 Hz. The system recorded the positions of markers, which were attached to heel, toe, malleolus, tibia, knee and femur of both legs as well as the sacrum, head and shoulders. Moreover, a cluster of 3 markers was attached to the iliac crest of both legs and 3 markers were attached to the platform. From the recorded marker positions, the movements of the CoM with respect to the ankle were reconstructed.

A custom built force plate was embedded in the moving platform and measured the forces and torques at a frequency of 360 Hz below each foot with a separate force sensor below the forefoot and heel (ATI-Mini45-SI-580-20). From the measured forces and torques, the position of the CoP with respect to the ankle position was calculated and subsequently used to calculate the generated ankle torque in each ankle.

The body sway and the ankle torques of the paretic and non-paretic limbs were used to assess the stabilizing mechanisms of the two limbs by using system identification techniques. The stabilizing mechanisms indicate the generated corrective torque in each ankle in response to a deviation of erect stance (body sway). The Dynamic Balance Contribution (DBC) can be deduced from the stabilizing mechanism, by calculating the ratio between the stabilizing mechanisms of the paretic and non-paretic leg to the stabilizing mechanism of both legs together. In short, the DBC is expressed as a fraction indicating how much of the required torque is generated in the paretic and non paretic leg in effort to stabilize the body in response to the platform perturbation.

4.3.4 **clinical tests**

The clinical tests were used to evaluate the functional balance and walking abilities. The principal investigator (EvA) conducted the following tests: the Berg Balance Scale (BBS) (Berg, 1989), the Timed Balance Test (TBT) (Bohannon *et al.*, 1984), the Timed Up and Go test (TUG) (Podsiadlo & Richardson, 1991), the Functional Ambulation Categories (FAC) (Holden *et al.*, 1984) and walking speed.

The BBS assesses the ability of the subject to maintain balance on a five point scale (0-4) during the performance of 14 common balance tasks, such as transfers, standing in different postures, reaching, turning. Higher scores reflect increased speed or ease of task-performance, and therefore indicate better balance. The BBS has been shown to yield data that have validity for stroke patients (Berg *et al.*, 1992a; Berg *et al.*, 1992b; Liston & Brouwer, 1996; Stevenson & Garland, 1996). The TUG test assesses the functional mobility by measuring the time to stand up from an armchair, walk a distance of 3 m, turn, and walk back to the chair and sit down again. The subjects performed 3 trials and the score was the average time of the three trials. The TBT assesses the balance ability on an ordinal 5 point scale. The subjects are given one point for each of 5 standing postures which they maintain for one minute. The standing postures have a progressively diminishing base of support. The FAC (0 to 5) distinguishes 6 levels of walking ability on the basis of the amount of physical support and supervision required to ambulate safely. The higher the level, the less support is necessary. The maximal and preferred gait velocity were measured as the subject walked 10 m over ground. The subjects performed 3 trials for each velocity and the average velocity was calculated in meters per second over the 3 trials. The subjects were allowed to use a cane during the test. Subjects used the same assistive device throughout all clinical tests on the different days.

4.3.5 **intervention**

The intervention consisted of 12 training sessions divided over 3 weeks (see Figure 4.1). Two training sessions were performed on the same day divided by a rest period of 45 minutes. During a training session, subjects stood on the movable platform wearing the safety harness suspended from the ceiling, while their balance was continuously challenged by horizontal movements of the platform in trials of 30 s. The direction of the horizontal movements

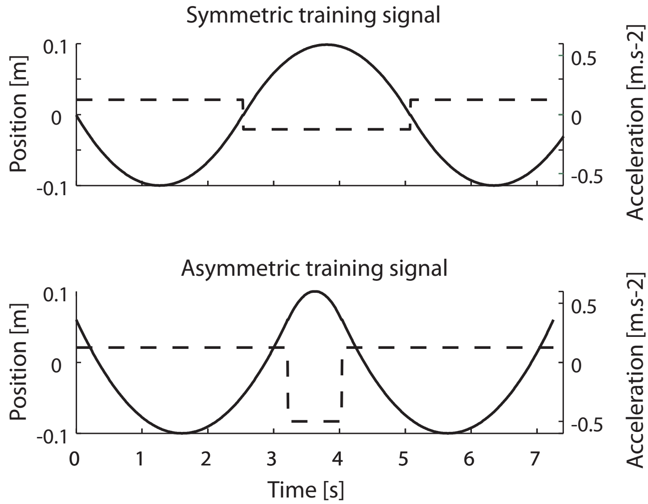

Figure 4.2. Design of the perturbation signal. The upper panel shows the symmetric signal with equal acceleration pulses in both directions. The lower panel shows what happens to the position signal when one of the acceleration pulses is increased in magnitude.

alternated between trials in anteroposterior direction and in mediolateral direction. Between trials subjects were given ample time to rest. On average the subjects performed 15 trials of platform perturbations.

The base of the perturbation signal consisted of a symmetric movement obtained by double integration of a pulse like acceleration pattern with pulse heights of equal magnitude (see Figure 4.2). The peak-to-peak amplitude of the perturbation was 0.2 m and the cycle time approximately 5 s. For support base movements the magnitude of the acceleration determines the magnitude of the perturbation. The acceleration of the platform can be thought of as being proportional to a force pushing at the CoM. When patients were well able to withstand the symmetric platform perturbation, the level of difficulty was adapted by increasing the height of one of the accelerations pulses. In order to keep the movement amplitude constant, the pulse durations were also adapted. The resulting platform movement (see for an example Figure 4.2) destabilized the body as it was pushed in the direction of the deceleration. We chose to increase only one of the pulses, since increasing both pulses would result in a destabilizing force, which subsequently was counteracted by stabilizing force in the opposite direction. By increasing only one pulse the subject was destabilized with an increasing force while the stabilizing force was kept constant, thereby increasing the demand on the postural control of the subject.

The use of the paretic leg in counteracting the perturbations was emphasized by both the design of the perturbation signal and the instructions of the therapist. The pulses for the anteroposterior and mediolateral direction were adapted in such a way that the patients were "pushed" toward their paretic side and backward respectively. The push toward the paretic side forced the patient to use their paretic leg in the balance response. Perturbation in the anteroposterior direction was intended to recruit automatic responses in the paretic leg to prevent a backward loss of balance. In both conditions, encouragement and instruction were also given by the physical therapist.

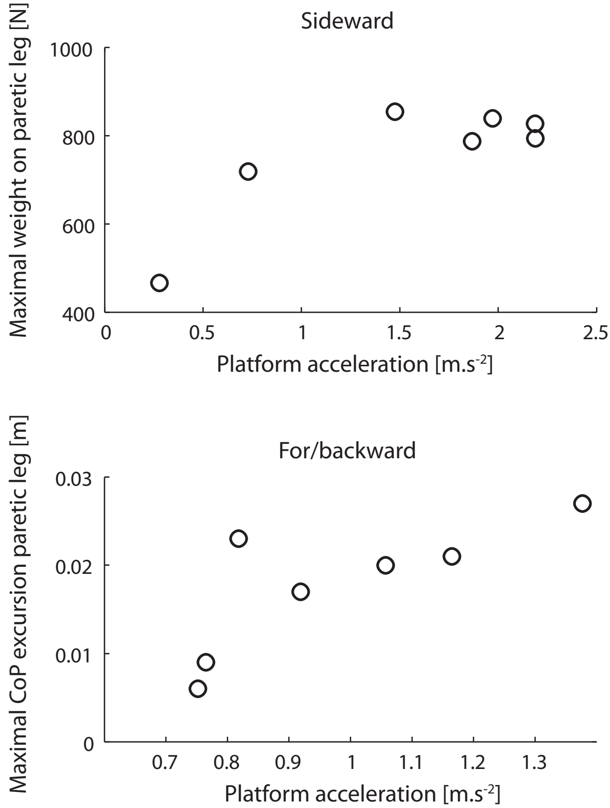

Figure 4.3. Example of the dependency of the weight bearing on the paretic leg (upper panel) and the maximal paretic CoP excursion (lower panel) on the magnitude of pulse acceleration in sideward and for/backward direction respectively. The different points are all from one training session and each point indicates a combination of values obtained from one single trial.

The height of the pulses was adapted in between trials and could differ between anteroposterior and mediolateral direction. Every training session started with a trial of symmetric perturbation signal and subsequently the level of difficulty was increased in 1 or 2 steps to the level reached at the end of the previous training session.

In order to evaluate whether the training actually increased the load of the paretic leg, we measured the ground reaction forces and CoP excursions below each foot using the force plate embedded in the movable platform. Although only inverse dynamics can provide an accurate estimation of the load on the paretic leg, we derived an indication of the loading from the vertical force below the paretic foot in the sideward direction, and the point of application in forward/backward direction. For the sideward movements we calculated the averaged maximal weight bearing on the paretic leg for each trial, by averaging the maximum vertical force below the paretic foot in response to each pulse that "pushed" the body toward the paretic leg. For the forward/backward perturbations we calculated the average maximal CoP excursion below the paretic foot for each trial, by averaging the maximal CoP excursions in response to all pulses that pushed the body backward.

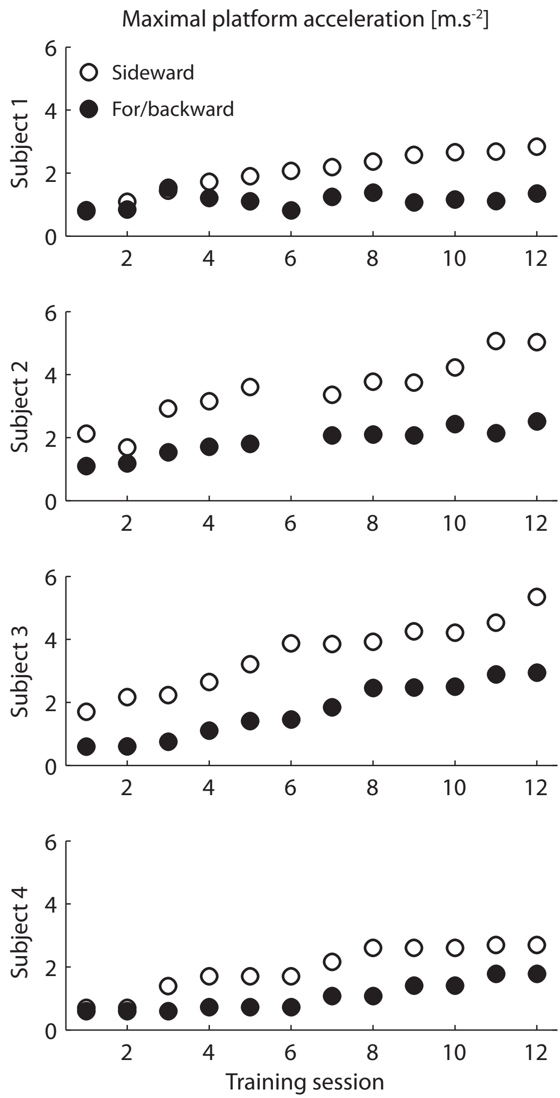

Figure 4.4. Magnitude of the maximal acceleration of the platform the subjects could withstand in every training session for the sideward (open circle) and for/backward direction (closed circle).

4.3.6 **analyses**

In single subject designs the data analyses is based on visual inspection of a graphical representation of the repeated measurements of the different variables. An effect of the intervention is demonstrated if the baseline changes with the introduction of the intervention, irrespective of the length of the baseline. A change can occur in the level, in the slope or in the variability of the measurements in the training phase with respect to the baseline phase (Kazdin, 1982).

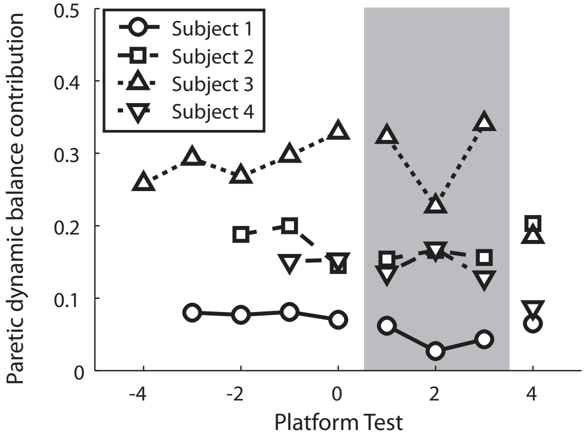

Figure 4.5. Dynamic balance contribution of the paretic leg for each of the subjects (separate line style). The training period is indicated with the gray area. The negative session numbers during the baseline period indicate the number of measurements before the training starts. The single measurement on the right side of the gray area shows the follow up score. The platform sessions are numbered independently from the clinical tests.

4.4 results

4.4.1 evaluation of training

By manipulating the size of the perturbations we aimed at increasing the response of the paretic leg to the perturbations. Figure 4.3 shows the averaged maximal CoP excursion below the paretic foot for the for/backward perturbations and the maximal paretic weight bearing for the sideward perturbations for all the trials in a training session of a representative subject. The magnitude of acceleration pulses was gradually increased over the different trials in both directions. For the sideward perturbations, the increase in acceleration pulse height was accompanied by an increase of the maximal weight bearing of the paretic leg. The increase in weight bearing indicated that the subject had to rely more on his paretic leg to restore balance. For the forward/backward perturbations, the increase in acceleration pulse height was accompanied with an increase in the CoP excursion, which reflected an increase in the generated torque in the ankle to counteract the perturbation.

To challenge the subjects, the magnitude of the acceleration was increased between trials to the maximum acceleration the subject could withstand for that moment. This maximum platform acceleration increased over the different sessions in both directions for all subjects (see Figure 4.4). The increase in platform acceleration differed between direction of movements as well as between subjects. The increase was smaller for forward/backward perturbations than for sideward perturbations and subjects 1 and 4 showed an apparent smaller increase for both directions than subjects 2 and 3.

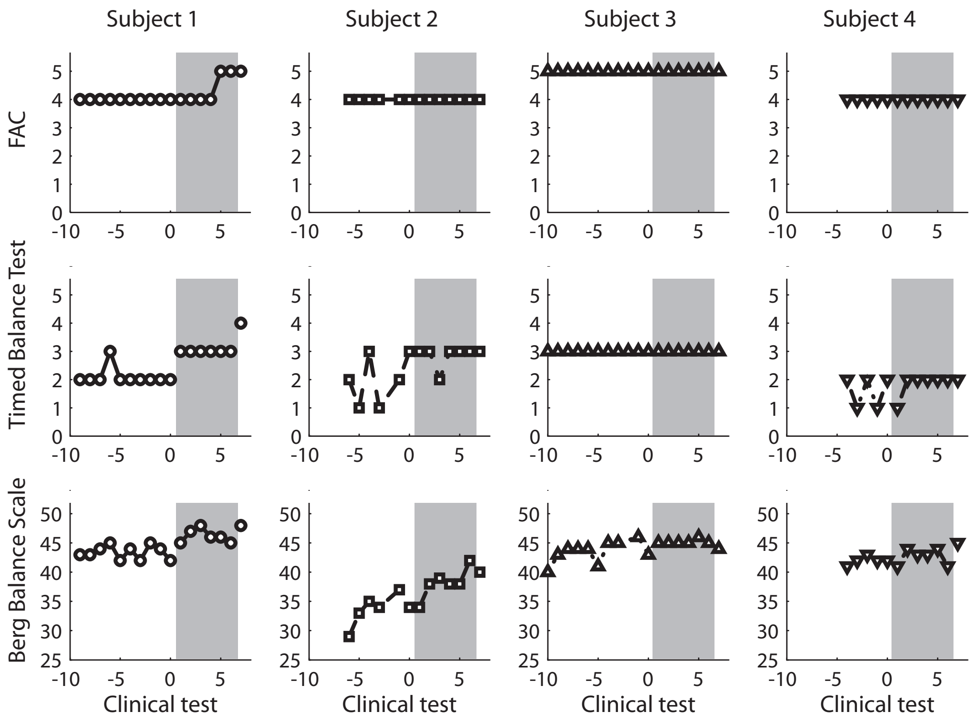

Figure 4.6. Scores on the Functional Ambulation Catagories (FAC), Timed Balance Test (TBT) and Berg Balance Score (BBS) for each subject (separate columns). The training period is indicated with the gray area. The negative session numbers during the baseline period indicate the number of measurements before the training starts. The single measurement on the right side of the gray area shows the follow up score.

4.4.2 **platform tests**

Platform tests were used to assess whether an increased control in the paretic leg could explain the increased ability to withstand perturbations. Although, the number of platform tests during the baseline and especially during the training period was relatively small to observe clear changes in trend or variation, the intervention generally did not show an effect on the contribution of the paretic leg to balance control (DBC)(see Figure 4.5). The largest changes in level occurred in the follow-up test. Patient 3 showed a large decrease of the DBC which indicated that the paretic leg contributed less to balance control. This decrease was comparable to the decrease observed between session 1 and 2 during the training period. Patient 4 also showed a decrease of the DBC which was outside the fluctuations seen in the training and baseline sessions.

4.4.3 **clinical tests**

Visual inspection of the data of the FAC and TBT (see Figure 4.6) showed that subject 1 showed an increase of the level of the FAC and the TBT during the intervention. Patient 2 and patient 4 maintained a score on the TBT during the training phase, which was only reached in some of the measurements during the baseline phase. For the BBS subject 1 and 2 showed an increase during the training phase with respect to the baseline phase. However for patient 2 the observed slope in the BBS scores was a continuation of the slope observed in the scores of the baseline phase. The observed improvements on the aforementioned scales were maintained during the follow up phase. Patient 1 even showed a further improvement on the TBT in the follow-up evaluation.

No clear effect of the intervention on the preferred or the maximal walking speed was observed (Figure 4.7). Although patient 1 showed an increase of maximal and preferred walking speed with respect to the start of the baseline period, the increase of the speed was already observed during the baseline period and continued or leveled off during the training period for the maximal and preferred speed respectively. For the TUG test, patient 2 showed a remarkable improvement at the start of the baseline period, which could probably be attributed to a training effect on the concerned test. Although this subject also showed a decrease of the TUG during the training period the observed improvements were small with respect to the improvements during the first tests. Subject 3 showed a gradual decline for the TUG which started during baseline and continued during the training phase. Subject 1 showed a clear stabilization on the test scores on the lowest level scored during the baseline period.

4.5 discussion

In this study we assessed the improvements of balance control of chronic stroke patients as a result of a specifically designed training consisting of platform perturbations that loaded the paretic leg. The results showed an increase of the performance on the trained task. All subjects could withstand larger perturbations in for/backward and sideward direction during the course of the training. The evaluation of the training showed that larger perturbations were accompanied by a larger demand on the paretic leg. However, the results of the platform test did not provide evidence for an increased contribution of the paretic leg to balance control. None of the patients showed a clear change in the dynamic balance contribution, which quantifies the contribution of the paretic leg in controlling the CoM. An increased ability to withstand perturbations without an accompanying increase of control in the paretic leg indicates that subjects learned to use their non paretic leg and trunk more effectively in counteracting the perturbations. That is, the subjects learned a compensatory strategy.

Apart of this study, the contributions of the paretic and non paretic leg during the course of balance and gait training are only assessed in a few other studies (Kirker *et al.*, 2000; de Haart *et al.*, 2004; Marigold *et al.*, 2005). However, these measurements are much needed to deduce whether restitution in the paretic leg or compensation of the non paretic leg is the main contributor to improvements. Evidence for restitution in the paretic leg was shown in two studies. In response to sideward external perturbations acute stroke subjects restored use of their paretic hip abductors (Kirker *et al.*, 2000). And in chronic stroke subjects, a decrease of the latency of the paretic rectus femoris, biceps femoris, tibialis anterior and gastrocnemius medialis was observed following balance training(Marigold *et al.*, 2005). In contrast, evidence for compensation of the non paretic limb was found in a study of de Haart and colleagues (2004). Using measures of a force plate below each foot, they found an increase in balance control while the regulating activity of the paretic leg did not increase. More studies are needed to elucidate the importance of the different recovery mechanisms in the increase in function of acute as well as chronic stroke patients.

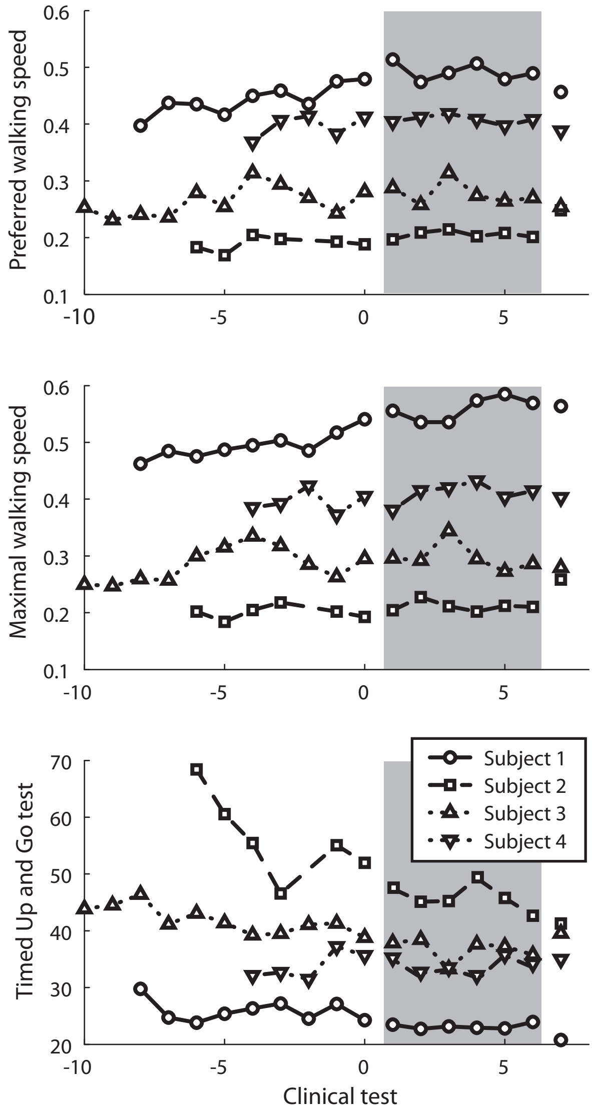

Figure 4.7. Preferred and maximal walking speed and time to complete the Timed Up and Go (TUG) test for each of the subjects (separate line style). The training period is indicated with the gray area. The negative session numbers during the baseline period indicate the number of measurements before the training starts. The single measurement on the right side of the gray area shows the follow up score.

transfer of training

This study showed limited transfer of training to more functional balance task. Only one subject (subject 1) exhibited a clear improvement in the clinical tests (Berg Balance Scale, Timed Balance Test, Functional Ambulation Categories). In the other subjects some changes in the clinical scores were observed during training. However, these changes were either stabilization of scores which were occasionally reached during the baseline or continuations of trends which were already observed during baseline. As a consequence these improvements could not be ascribed solely to the balance training. We hypothesized that a partial restitution of function in the paretic leg would have a beneficial effect on functional balance. The observed limited transfer might be explained from the fact that improvements during training were ascribed to learning a compensation strategy instead of restitution in the paretic leg. This strategy might be effective in the specific situation of the training but might be of little use in other situations. In other words, a strategy to counteract external perturbations might not be appropriate for counteracting internal perturbations induced by self generating motions like rising from a chair, walking and reaching.

The association between withstanding internally produced and externally applied perturbations is far from clear. To our knowledge, so far no study has specifically addressed this association. Also the relation between counteracting external perturbations and scores on clinical test should be further assessed. Recently, Vearrier and colleagues (2005) showed that an intensive massed balance training resulted in a average increase of functional balance which was accompanied by improvements in counteracting perturbations, reflected in a decrease of the time needed to restabilize after a platform perturbation. However, three of the patients in this study showed no change in clinical scores although their time to restabilize decreased by at least 30 %. These latter results are in agreement with the results of our study and show the necessity to further explore the relation between the reactive balance control and functional balance.

methodological considerations

The fact that we did not find a clear effect of the training might have been due to the design of the study and the design of the training. The number of training sessions was relatively low compared to other studies (Walker *et al.*, 2000; Marigold *et al.*, 2005). However, since two subjects (3 and 4) exhibited hardly any change in scores during the training, nor any improvement toward the end of the training, it is not likely that a longer training period would have resulted in improvements. Increasing the training duration is no guarantee for increasing the training effectiveness. Even after an intensive massed practice intervention (Vearrier *et al.*, 2005) consisting of training six hours a day for two weeks, three out of ten patients did not show any improvement on the clinical scales. Probably these patients like the two patients in our study were not susceptible for improvements. Their functional performance might be limited by impairments which are not easily corrected by a training intervention. Very limited information is available about which characteristics/impairments of a patient limit the efficacy of training for the particular patient. It is of great importance that these characteristics will be identified as they are required to tailor the training programs to each specific patient.

The use of a multiple baseline design might also have obscured the positive effects of the training. In this design a large number of baseline assessments is prescribed to carefully assess the baseline performance of the patients. However, different patients showed an increased performance on one or more of clinical tests, instead of the expected variation of the scores around a mean value. This increase might be caused by a "learning effect" or a "fitness effect". Despite that all used tests have a high test-retest reliability, the uncommon high number of assessments could have led to learning to perform (specific aspects) of the administered tests. Subjects could have tried different strategies to perform the test and as the time between assessments was small (frequency was twice a week), subjects might have remembered the best strategy from the previous assessment. Learning of specific aspect might have been further promoted as specific capabilities of the subject were assessed in different tests, leading to an increased amount of practice. For example the capacity of the subject to turn around was part of the Timed Up and Go test and was one of the tasks scored in the BBS. The "fitness effect" constitutes of increased fitness level of the subjects due to repeated assessment of the tests. The duration of the administration of all clinical tests, took about 45 to 60 minutes, depending on the amount of rest the subjects needed in between. All tests consisted of performing functional tasks, like walking, turning, performing transfers, reaching. As a consequence, performing all the test could be regarded as a task specific training session on its own. These "training sessions" could have led to their own effects, like increased fitness level and balance control.

The increase in performance during baseline interfered with assessing the efficacy of the platform training per se. Still, an additional effect of the training could have been expressed by a larger increase of the scores during training. However, the clinical tests showing a trend during baseline, did either show a continuation of this trend, or leveling off the trend. Trends in baseline data have been observed more often in studies using single case research designs and if not recognized as such can result in false conclusions about effects of the training (Marklund & Klassbo, 2006).

Assuming that chronic stroke patients have limited capacity for intervention-related improvement, the improvements during baseline, may have limited the degree of change possible with training. Therefore, to prevent the occurrence of a fitness and training effect in future studies using single case designs, the number of different clinical tests to assess the effect of training should be minimized so the collection of tests is not a training on its own and so there is no transfer of learning specific capabilities from one test to the other. This especially applies for studies involving subjects with low functional capabilities as the subject with the lowest scores on the different tests (subject 2) showed the largest improvements during baseline.

The current study has some methodological limitations. First, the first author conducted the testing and was also involved in the training of the subjects. As such he was not blinded for the onset of the training period. Second, the follow up only consisted of one clinical test and one platform test. During the baseline period considerable variation was observed in the clinical scores. This variation can also be expected, when performing multiple assessments during follow up. Subject 1 showed a remarkable increase in performance on the Timed Balance Test and the Timed Up and Go test, whereas subject 2 showed a remarkable increase in the walking velocity. We could have determined with more certainty whether these follow up effects were true or caused by chance, if we had conducted a more elaborated follow up.

The contribution of the paretic leg to balance control was only evaluated for perturbations in forward/backward directions and not for sideward perturbations. De Haart and colleagues (De Haart *et al.*, 2004) showed that frontal balance improved to a greater extent than sagittal balance in a group of 37 in-patients during the course of rehabilitation. They also showed that this improvement was not accompanied by a more symmetric distribution of the regulating activity below the paretic and non paretic foot. However, the latter measurements were based on CoP excursions, and these only reflect the generated ankle torques. Especially for sideward perturbations, torques generated at the hips, play a crucial role in balance control (Rietdyk *et al.*, 1999). At this moment, no method is available yet that can be used to objectively quantify the contribution of the ankle and hip joint torques to the stabilization of the CoM in the frontal plane. We are currently working on extending the methods to determine the DBC in the frontal plane.

4.5.1 evaluation of the training

In the balance training the paretic leg was elicited to respond to the perturbations. By varying the height of the acceleration pulse, the load on the paretic leg could be increased, which resulted in a "forced use" of the paretic leg. Forced use is one of the key elements, the other being massed practice (approximately 6 hours a day for 2 weeks), in Constraint Induced Movement Therapy (CIMT). CIMT is a therapy for the upper extremities in which the patients are required to use their paretic arm for functional activities by restricting the use of unaffected arm. CIMT has been shown to result in a significant and clinically relevant increase in arm motor function (Wolf *et al.*, 2006). The success of CIMT for the upper extremities has led to the development of variations of this therapy for the lower extremities (Vearrier *et al.*, 2005; Marklund & Klassbo, 2006). However, constraining the use of the non paretic leg is problematic as generally both legs are necessary to execute tasks such as walking or balance control. Marklund and Klassbo (2006) used a plastic whole-leg orthosis to immobilize the non paretic leg. We believe that this is an inappropriate method, as it can lead to very unsafe situations and furthermore walking with a stiff non paretic leg is a fundamentally different task than normal walking. In the intervention of Vearrier and colleagues (Vearrier *et al.*, 2005) therapist emphasized the use of the paretic leg. In addition, subjects received auditory and visual feedback on the symmetry of their performance. Although the feedback makes the patient more aware of the contribution of his paretic leg it does not force the subject to use it. The platform perturbations used in this study might be a valuable addition to the massed practice therapies as it allows for a safe implementation of "forced use" which can be easily adapted to the capabilities of the patient.

We chose to manipulate only the height of the acceleration pulse in one direction and adapted the duration of the acceleration pulses to keep the height in the other directions as well as the range of the platform movement constant. Because of the necessity to keep the movement range constant, an increase of the pulse height was accompanied by a decrease of the duration of the pulse (see Figure 4.2). This decrease of duration, limited the increase of the impulse (integration of area below the pulse). The adaptation of the pulse height was adequate to challenge the subjects in our study. However when training subjects who have learned compensatory balance control, manipulation of pulse height in combination with the pulse duration could be necessary to challenge the subject to a sufficient degree. In that case patients cannot only be "pushed" with a larger force to their paretic side, but also for a longer time.

4.5.2 conclusion

In this study we only find minor evidence for a beneficial effect of balance training based on platform movements in chronic stroke survivors. By manipulation of the platform acceleration we were able increase the demand on the paretic leg. The increased demand was not a strong enough trigger to improve control in the paretic leg for the stroke subjects in this study.

However, for acute patients the platform perturbations might induce restitution of the paretic leg, since these patients have not yet learned compensatory strategies with the stronger limb. Furthermore, experiencing perturbations under controlled and safe circumstances could have a beneficial effect on the patients' confidence in performing activities without fear of falling. Especially acute patients do not know yet where they are still capable of. In that case, a low self-confidence could hinder patients in performing tasks to an equal extent as their physical impairments. Future studies should be designed to investigate the effect of surface perturbations for training of acute stroke patients.

4.6 references

Aruin AS, Hanke T, Chaudhuri G, Harvey R & Rao N. (2000). Compelled weightbearing in persons with hemiparesis following stroke: the effect of a lift insert and goal-directed balance exercise. *Journal of Rehabilitation Research and Development* 37: 65-72

Backman CL & Harris SR. (1999). Case studies, single-subject research, and N of 1 randomized trials: comparisons and contrasts. *American Journal of Physical Medicine & Rehabilitation* 78: 170-176

Backman CL, Harris SR, Chisholm JA & Monette AD. (1997). Single-subject research in rehabilitation: a review of studies using AB, withdrawal, multiple baseline, and alternating treatments designs. *Archives of Physical Medicine and Rehabilitation* 78: 1145-1153

Badke MB & Duncan PW. (1983). Patterns of rapid motor responses during postural adjustments when standing in healthy subjects and hemiplegic patients. *Physical Therapy* 63: 13-20.

Barclay-Goddard R, Stevenson T, Poluha W, Moffatt ME & Taback SP. (2004). Force platform feedback for standing balance training after stroke. *Cochrane Database of Systematic Reviews* CD004129

Berg K. (1989). Measuring balance in the elderly: A Review. *Physiotherapy Canada* 41: 304-311

Berg KO, Maki BE, Williams JI, Holliday PJ & Wood-Dauphinee SL. (1992a). Clinical and laboratory measures of postural balance in an elderly population. *Archives of Physical Medicine and Rehabilitation* 73: 1073-1080

Berg KO, Wood-Dauphinee SL, Williams JI & Maki B. (1992b). Measuring balance in the elderly: validation of an instrument. *Canadian Journal of Public Health* 83 Suppl 2: S7-11

Bohannon RW, Larkin PA, Cook AC, Gear J & Singer J. (1984). Decrease in timed balance test scores with aging. *Physical Therapy* 64: 1067-1070

De Haart M, Geurts AC, Dault MC, Nienhuis B & Duysens J. (2005). Restoration of weight-shifting capacity in patients with postacute stroke: a rehabilitation cohort study. *Archives of Physical Medicine and Rehabilitation* 86: 755-762

de Haart M, Geurts AC, Huidekoper SC, Fasotti L & van Limbeek J. (2004). Recovery of standing balance in postacute stroke patients: a rehabilitation cohort study. *Archives of Physical Medicine and Rehabilitation* 85: 886-895

Diener HC, Bacher M, Guschlbauer B, Thomas C & Dichgans J. (1993). The coordination of posture and voluntary movement in patients with hemiparesis. *Journal of Neurology* 240: 161-167

Hocherman S, Dickstein R & Pillar T. (1984). Platform training and postural stability in hemiplegia. *Archives of Physical Medicine and Rehabilitation* 65: 588-592

Holden MK, Gill KM, Magliozzi MR, Nathan J & Piehl-Baker L. (1984). Clinical gait assessment in the neurologically impaired. Reliability and meaningfulness. *Physical Therapy* 64: 35-40

Holt RR, Simpson D, Jenner JR, Kirker SG & Wing AM. (2000). Ground reaction force after a sideways push as a measure of balance in recovery from stroke. *Clinical Rehabilitation* 14: 88-95

Hyndman D, Ashburn A & Stack E. (2002). Fall events among people with stroke living in the community: circumstances of falls and characteristics of fallers. *Archives of Physical Medicine and Rehabilitation* 83: 165-170

Ikai T, Kamikubo T, Takehara I, Nishi M & Miyano S. (2003). Dynamic postural control in patients with hemiparesis. *American Journal of Physical Medicine & Rehabilitation* 82: 463-469

Jorgensen L, Engstad T & Jacobsen BK. (2002). Higher incidence of falls in long-term stroke survivors than in population controls: depressive symptoms predict falls after stroke. *Stroke* 33: 542-547

Kazdin AE. (1982). *Single-case research desings*. Oxford University Press, New York.

Kirker SG, Jenner JR, Simpson DS & Wing AM. (2000). Changing patterns of postural hip muscle activity during recovery from stroke. *Clinical Rehabilitation* 14: 618-626.

Liston RA & Brouwer BJ. (1996). Reliability and validity of measures obtained from stroke patients using the Balance Master. *Archives of Physical Medicine and Rehabilitation* 77: 425-430

Marigold DS & Eng JJ. (2006). Altered timing of postural reflexes contributes to falling in persons with chronic stroke. *Experimental Brain Research* 171: 459-468

Marigold DS, Eng JJ, Dawson AS, Inglis JT, Harris JE & Gylfadottir S. (2005). Exercise leads to faster postural reflexes, improved balance and mobility, and fewer falls in older persons with chronic stroke. *Journal of the American Geriatrics Society* 53: 416-423

Marklund I & Klassbo M. (2006). Effects of lower limb intensive mass practice in poststroke patients: single-subject experimental design with long-term follow-up. *Clinical Rehabilitation* 20: 568-576

Nardone A, Galante M, Lucas B & Schieppati M. (2001). Stance control is not affected by paresis and reflex hyperexcitability: the case of spastic patients. *Journal of Neurology, Neurosurgery & Psychiatry* 70: 635-643.

Nyberg L & Gustafson Y. (1997). Fall prediction index for patients in stroke rehabilitation. *Stroke* 28: 716-721

Ottenbacher KJ. (1986). *Evaluating Clinical change. Strategies for occupational and physical therapists*. Williams & Wilkins, Baltimore.

Podsiadlo D & Richardson S. (1991). The timed "Up & Go": a test of basic functional mobility for frail elderly persons. *Journal of the American Geriatrics Society* 39: 142-148

Rietdyk S, Patla AE, Winter DA, Ishac MG & Little CE. (1999). NACOB presentation CSB New Investigator Award. Balance recovery from medio-lateral perturbations of the upper body during standing. North American Congress on Biomechanics. *Journal of Biomechanics* 32: 1149-1158

Sackley CM. (1991). Falls, sway, and symmetry of weight-bearing after stroke. *International Disability Studies* 13: 1-4

Shumway-Cook A, Anson D & Haller S. (1988). Postural sway biofeedback: its effect on reestablishing stance stability in hemiplegic patients. *Archives of Physical Medicine and Rehabilitation* 69: 395-400.

Stevenson TJ & Garland SJ. (1996). Standing balance during internally produced perturbations in subjects with hemiplegia: validation of the balance scale. *Archives of Physical Medicine and Rehabilitation* 77: 656-662

Van Asseldonk EH, Buurke JH, Bloem BR, Renzenbrink GJ, Nene AV, van der Helm FC & van der Kooij H. (2006). Disentangling the contribution of the paretic and non-paretic ankle to balance control in stroke patients. *Experimental Neurology* 201: 441-451

Vearrier LA, Langan J, Shumway-Cook A & Woollacott M. (2005). An intensive massed practice approach to retraining balance post-stroke. *Gait & Posture* 22: 154-163

Walker C, Brouwer BJ & Culham EG. (2000). Use of visual feedback in retraining balance following acute stroke. *Physical Therapy* 80: 886-895.

Winstein CJ, Gardner ER, McNeal DR, Barto PS & Nicholson DE. (1989). Standing balance training: effect on balance and locomotion in hemiparetic adults. *Archives of Physical Medicine and Rehabilitation* 70: 755-762.

Wolf SL, Winstein CJ, Miller JP, Taub E, Uswatte G, Morris D, Giuliani C, Light KE & Nichols-Larsen D. (2006). Effect of constraint-induced movement therapy on upper extremity function 3 to 9 months after stroke: the EXCITE randomized clinical trial. *JAMA* 296: 2095-2104

chapter 5

influence of guiding force fields on visuomotor learning

submitted

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5.1 abstract

In (re)learning of movements, guidance can be used to direct the adaptations that occur between the different attempts. However, guidance will decrease the magnitude of the execution errors, considered to be the main driving force in adaptation. During guidance, interactions occur which are also known to drive adaptation. The aim of this study was to assess how external guidance affects motor adaptation. We hypothesized that adaptation based on interaction forces would be less efficient than adaptation based on the execution errors. Five groups of subjects adapted to a visual rotation task, while being guided by per group different force fields. The force fields differed in magnitude and direction, in order to discern the adaptation based on execution errors and interaction forces. The execution error did indeed play a key role in adaptation; the more the guiding forces restricted the occurrence of execution errors, the smaller the amount and rate of adaptation. However, the force field that enlarged the execution errors did not result in an increased rate of adaptation. The presence of a small amount of adaptation in the groups who did not experience execution errors during training suggested that adaptation could be driven on a much slower rate and/or to a lesser extent, based on minimization of the muscular effort as was evidenced by a slight gradual decrease of the interaction forces during training. These results extend the recent identified role of optimizing muscular effort in adaptation to a dynamic force field, to adaptation to a kinematic perturbation.

5.2 introduction

Guidance of movements can be used to demonstrate how a movement should be performed. As such, the guidance is used for learning new skills in sports, but also for relearning motor control after having a stroke (Kahn *et al.*, 2006). In the latter case, the guidance was traditionally applied manually by a therapist. However, in the last decade different robotic devices (Colombo *et al.*, 2000; Lum *et al.*, 2002; Ferraro *et al.*, 2003; Hogan & Krebs, 2004) have been developed, which can provide unlimited guidance during the recommended highly repetitive practicing (Kwakkel *et al.*, 2004; Teasell *et al.*, 2005).

The applied guidance should direct the adaptations that occur during practicing, such that eventually the goal behavior is learned. The process of motor learning has been thoroughly studied by exposing subjects to a specific perturbation, and quantifying the trial to trial adaptations that occur to compensate for the perturbation and return to the normal movement patterns (Thoroughman & Shadmehr, 1999, 2000). The applied perturbations can be divided in kinematic and in dynamic perturbations. In kinematic perturbations, the perturbations are induced by a distortion of the visual feedback of the hand and /or arm perturbations (Krakauer *et al.*, 1999; Ghez *et al.*, 2000; Krakauer *et al.*, 2000; Sainburg & Wang, 2002; Tong *et al.*, 2002; Caithness *et al.*, 2004; Wang & Sainburg, 2005), whereas in dynamic perturbations, a mechanical disturbance is applied that for instance simulates a viscous or a spring load (Shadmehr & Mussa-Ivaldi, 1994; Goodbody & Wolpert, 1998; Donchin *et al.*, 2003; Scheidt *et al.*, 2005). Generally, the introduced distortions change the interaction of the arm in the environment, requiring a new sensorimotor mapping (often referred to as an internal model) to be formed between the sensed state and the required motor commands to reach the target. Different studies have indicated that the magnitude of the errors during execution of the movement is the main driving force in adaptation and as such in updating of these internal models (Thoroughman & Shadmehr, 2000; Scheidt *et al.*, 2001; Pipereit *et al.*, 2006). This brings forward an intriguing issue for the application of guidance during adaptation of the movements. As guidance is generally aimed to reduce the execution errors, it would slow down learning in stead of accelerating it, unless other information obtained from guidance can be effectively used in adaptation.

Instead of using execution errors, subjects could also use dynamic criteria in adapting their movements, such as the exerted force on the limb or the muscular effort. In the adaptation to novel dynamic environments, minimization of muscular effort or forces have previously been shown to be involved (Scheidt *et al.*, 2000; Emken *et al.*, 2007b). Scheidt and colleagues (2000) used a simulated “haptic channel” that prevented the occurrence of kinematic after effects after removal of the previously learned viscous force field. They showed that subjects made movements while simultaneously exerting perpendicular forces to the haptic channel that were similar to the forces required to compensate for the viscous force field. Despite the absence of kinematic errors, subjects disadapted by decaying the forces exerted on the channel over the different movements. Still, the disadaptation occurred at a much slower rate than when kinematic errors were allowed to occur. Further evidence for a contribution of muscular effort in adaptation was recently provided by Emken and colleagues (2007b). They examined the adaptation to an externally applied force field during the swing phase of walking and showed that a model describing the temporal evolution of error (Thoroughman & Shadmehr, 2000; Scheidt *et al.*, 2001) could be derived from minimization of a cost function that is a weighted sum of the execution error and muscular effort.

Minimizing of muscular effort could possibly also be used in adaptation while receiving guidance. Generally, guidance is only applied when the movement deviates from the optimal trajectory and as such the muscular effort is higher than the task strictly requires. The experienced interaction forces can be regarded as a dissipation of muscular effort. Minimizing of these interaction forces would result in the most efficient movement and therefore in the required adaptation.

The aim of this study was to assess how external guidance affects motor adaptation. As guiding forces would interfere with the disturbing forces during adaptation to a novel dynamic environment (Emken *et al.*, 2007a), we applied different guiding force fields while subjects adapted to a kinematic distortion, that is, a visual rotation task. In addition to the control group in which subjects learned the visuomotor rotation without any additional forces, we had four groups exposed each to a different force field. All but one of these force fields applied forces only in the direction perpendicular to the target direction, which necessitates the subjects to move in the target direction themselves. In these force fields, the exerted force was dependent on the execution error during reaching. As a consequence subjects could have adapted to the visuomotor rotation by either minimizing the exerted forces or the execution errors. To discern the adaptation based on execution errors and forces, the force fields differed in magnitude and direction. Subjects in the error enhanced group received hand forces that were proportional and in the same direction as the execution errors, effectively enlarging the execution errors. In the soft and hard guidance groups, error correcting forces were applied to the hand which were proportional but opposite to the execution errors. In the soft guidance group, the low stiffness of the force field still allowed considerable execution errors. However, in the hard guidance group, the high stiffness formed a haptic tunnel, denying all but very small deviations (<1.5 mm) from the optimal trajectory. This channel was similar to the channel used by Scheidt and colleagues (2000). In the last group, the passive group, the subjects were moved along the optimal trajectory by the robot and were instructed not to intervene. Thus these subjects could adapt neither based on the execution error nor on the minimization of perpendicular interaction forces.

We hypothesized that like in adaptation to a dynamic force field, adaptation to a visuomotor task can be driven by dynamic criteria, however at a slower rate than adaptation based on the execution errors. In other words, we expected that force fields that restrained execution errors (soft guidance) would result in slower adaptation, whereas the force fields that increased the execution errors (error enhanced) would result in faster adaptation. Still, the minimization of interaction forces could solely drive adaptation in the absence of execution errors, as in the hard guidance group. In absence of execution error and muscular effort (passive) we expected that no adaptation would take place.

5.3 materials and methods

5.3.1 subjects

Fifty healthy subjects (age 20-50 yr, 16 female) were included, all giving their written informed consent prior to the experiment. The protocol was approved according to the institution's regulations. All subjects were right-handed, had no history of neurological impairments and had a normal or normal corrected vision. The subjects were randomly assigned to one of the following training programs, "Active" (A), "Passive" (P), "Hard Guidance" (HG), "Soft Guidance" (SG) and "Error Enhanced" (EE) training.

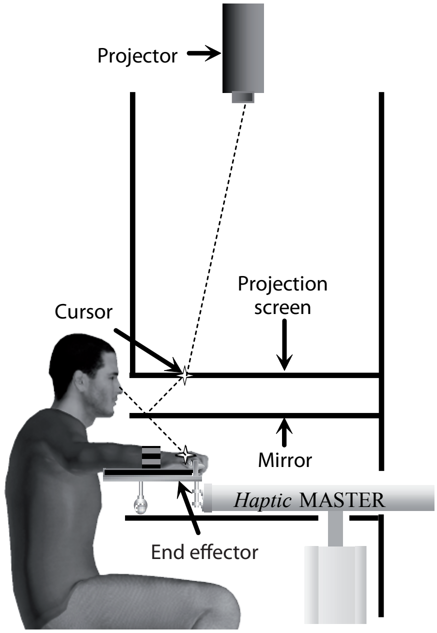

Figure 5. 1. Schematic overview of experimental setup. Subjects sat behind a closet-like box and held with their right hand the end-effector of a haptic robot. Subjects looked into a mirror just below shoulder level to a projection of their (rotated) right hand position and the targets. The mirror prevented sight of their right arm. The arm was supported by a surface through a mechanism that allowed horizontal movements with low friction.

5.3.2 experimental apparatus and recordings

The subjects were seated (see Figure 5.1) and made reaching movements in the horizontal plane with their right arm while the right hand was holding the "end-effector" of a 3D haptic robot, the HapticMASTER (Moog FCS, Nieuw-Vennep, The Netherlands), which we restricted to functioning in the horizontal 2D plane. The force exerted by the HapticMASTER on the hand was controlled at 2500 Hz to create the guiding forces described below in further detail. The arm robot was placed in a closet-like box. The subjects were instructed to look into a mirror to see a projection of their right hand position on a screen located parallel and just above the mirror. The combination of a mirror and projection screen gave the illusion that the projected image was in the same horizontal plane as the hand, resulting in a veridical projection. The mirror also prevented direct sight of the arm. The right-hand position was indicated with a 6 mm blue sphere, in the following referred to as "cursor". The targets were presented as yellow spheres with a 10 mm diameter. The visual scene was updated with a frequency of 100 Hz. The arm was supported against gravity by a support mechanism which allowed low friction movements over an underlying surface (see Figure 5.1). The arm support also prevented wrist movement. As a result, movements of the hand were restricted to the horizontal plane and solely the result of joint rotations around the vertical axes of elbow and shoulder. Velocity and position data of the end-effector of the HapticMASTER were sampled at 200 Hz.

5.3.3 procedure

Subjects made centre-out reaching movements with their right hand to one of five different targets equally spaced (72° apart) about the perimeter of a circle of 10 cm radius. The center of movements was always in the midsagittal plane 10 cm beneath right shoulder position. The starting posture was obtained by a shoulder plane of elevation rotation of 45° (Wu et al., 2005) and elbow flexion of 90°. At the start of each trial, a target was presented and a short "beep" triggered movement. The order of the targets was randomized in each cycle, where a cycle consisted of one trial to each target (five movements). Subjects received feedback about the accuracy of the movement duration by means of target color and a sound. The end of the movement was defined as the moment at which the velocity decreased below 5 mm/s. The target color changed into green when the cursor was inside the target within the prescribed time interval [510-690 ms] and a normal tone was heard. Too fast or too slow movements were accompanied by a change of the color of the target into blue and red respectively, and a dropping and refuting sound. During all stages, the number of accurate reaches was shown on the visual display in the right upper corner. Between movements, the cursor was made invisible while the arm was returned to the starting position by the robot.

Visual distortion of hand position, during training and extended training, was a 30° CCW rotation about the starting location of movements. When exposed to the visual rotation, subjects received different force fields, depending on the group to which they were assigned. In addition to these force fields, all groups except P also felt the inertial forces of a 2 kg virtual floating mass in the virtual environment created by the admittance controlled haptic robot.

- Group A (Active learning) did not receive any additional forces during training.
- Group P (Passive training) subjects were moved along the optimal trajectory to the target by stiff control (K=5000 N/m) of the robot. This optimal trajectory was defined as a straight line from centre to target position, with a bell-shaped velocity profile:

  $$v(t) = \frac{y_R}{T}\left(1 - \cos\left(2\pi\frac{t}{T}\right)\right), \quad 0 < t < T, \qquad (1)$$

  where T is the movement duration (T = 600 ms), t is time and y_R is the distance between centre and target. Subjects in this group were specifically instructed not to intervene with the applied robot, so not to assist or resist the imposed movements.
- Group HG (Hard Guided training) movements were restrained to the desired path by a stiff force field perpendicular to the optimal trajectory acting on the right hand. Since the stiffness along this optimal trajectory was zero, the subjects were free to control the progress along the path. Force F is a function of deviation from a straight path from the start to the target:

  $$F(x) = K \cdot x \qquad (2)$$

  with K = 5000 N/m. Cursor trajectory error x_C', same in magnitude as hand trajectory error x_C, was described as the distance between current cursor position and the y-axis, the optimal hand trajectory (see Figure 5.2). The relation between force and deviation is visible as a high gradient surface "V", shown in Figure 5.3a.
- Group SG (Soft Guided training) experienced a similar force field as group HG but with a smaller stiffness (K = 300 N/m). The low gradient gray surface in Figure 5.3a illustrates the soft guidance force-error relation.

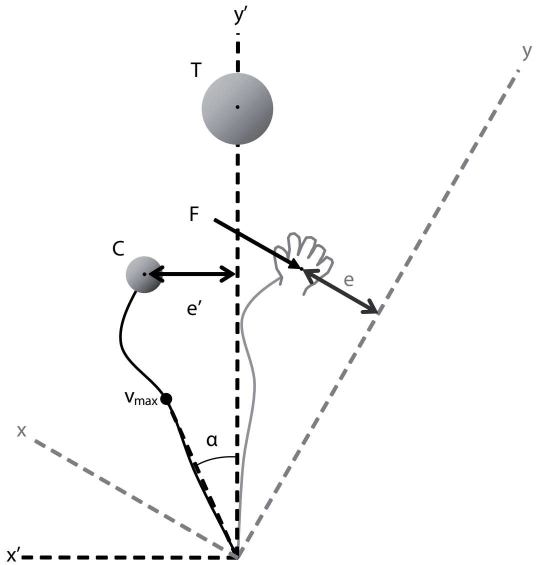

Figure 5.2. Definitions in the calculation of the guiding forces and the direction error. Error (e) is the distance x' from current cursor (small sphere) position to the y'-axis, which is also the optimal cursor trajectory from centre to target (large sphere). Force F, dependent on hand position error (e = e'), acts on the arm and is perpendicular to the y-axis and parallel to the x axis. The cursor path is 30° CCW rotated from the actual hand position. Symbol α represents the angle between cursor position at maximum velocity (vmax) and optimal trajectory.

- Group EE (Error Enhanced training) was exposed to an error enhancing force field (see Figure 5.3b). When subjects deviated from the optimal path, they experienced a force, which even pushed them farther away. In this case, F was described as

$$F(x,y)=\begin{cases} A(y_C)\left(\frac{1}{2}-\frac{1}{2}\cos\left(2\pi\frac{x}{b}\right)\right), & 0\leq x\leq\frac{1}{2}B \quad \frac{3}{2}B\leq x\leq 2B \\ 0, & x>2B \\ A(y_C), & \frac{1}{2}B<x<\frac{3}{2}B \end{cases} \tag{3}$$

where B = 0.05 m is a quarter of the area in which forces are present (See Figure 5.3 b). Factor $A(y)$, expressed as

$$A(y_C)=-K\cdot y_C \tag{4}$$

was a maximum force, dependent (K = 500 N/m) on current position y-value (y_C) , the distance between centre and the projection of hand position on the optimal path.

Each group attended a program that consisted of four different stages, in following order:

1. In the familiarization stage, the subject got familiar with the haptic and virtual environment. Participants executed 100 reaching movements, with every fifth movement a catch trial interspersed and the directions randomized per five movements. The last five trails were considered as baseline. During this stage, none of the groups had forces present.
2. During the training stage, participants performed the task in the visual rotated field. All subjects performed 60 cycles of five trials (300 movements). Every tenth movement a catch trial was interspersed to monitor the adaptation of subjects to the visuomotor rotation. As subjects in the passive group were not supposed to generated movements themselves during the training, the catch trials were preceded by an additional tone to make the subjects aware of when they were required to reach on their own. Generally, the adaptation to the visuomotor rotation is determined from the movement trajectory error (see Figure 5.2 for error definition) in subsequent training trials (Tong *et al.*, 2002). However, force fields influenced the magnitude of the error, which compromised error comparison of training trials in different groups. Therefore, we interspersed catch trials in which the group dependent force fields were turned off. Furthermore, during catch trials visual feedback was not presented to capture uncorrected movements and to rule out active learning moments, as reaching without additional forces with visual feedback would mimic the active learning condition.
3. The extended training stage deviated from the previous stage in that the catch trials occurred every fifth movement and that movements during these generalization catch trials were directed to generalization targets which deviated 36° from the trained directions, that is, located exactly between the original targets along the circle. Subjects performed 30 cycles of movement (150 movements), which included 30 generalization catch trials.
4. In the washout stage, visual feedback returned to "normal". All visual distortions and force fields, if any, were turned off. This was the unlearning phase. Hand position was visible during all movements, with a total of 100 movements (twenty cycles).

In total, subjects performed 130 cycles of five movements. With short time-outs between every stage, subjects spend approximately 36 minutes to perform the experiment.

5.3.4 **data analysis**

Movement position and velocity data of the hand were used to assess the reaching performance of the subjects in the different stages and conditions. We used the directional error as a measure of the execution error. Previous studies have shown that directional error is a sensitive and intuitive measure of adaptation to visuomotor rotations (Krakauer *et al.*, 2000; Sainburg & Wang, 2002). The directional error is calculated as the angle between the vector from the starting position to the cursor position at maximum velocity and the vector from the starting position to the target (see Figure 5.2). Furthermore the directional error captures the learning in feedforward control and is insensitive for contributions of feedback mechanism to the final portion of the reaching movement. Baseline performance was quantified as the mean directional error of the last cycle of reaching movements during the familiarization stage. The catch trials and the generalization catch trials were each divided into six blocks of five trials and mean values of every block were calculated. The after effect was assessed by calculating the mean value of the directional errors of the first cycle in the washout phase.

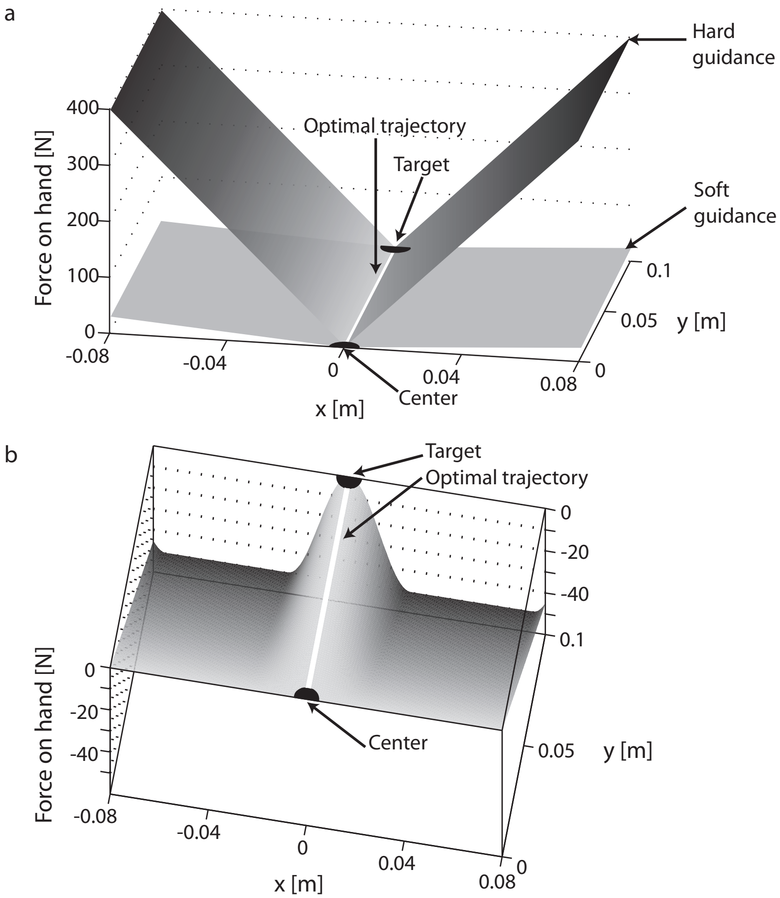

Figure 5.3. Forces exerted on the hand as a function of the deviation of the optimal trajectory (x) and the distance to the target (y). The forces are exerted in a direction perpendicular to the optimal trajectory. In (a) the forces for soft guidance (300 N/m) and hard guidance (5000 N/m) are depicted. These forces "pushed" the hand of the subject towards the optimal trajectory in order to decrease the reaching errors. (b) indicates the error enhancing forces, which were directed away from the optimal trajectory (indicated by the negative magnitude of the forces) to increase the reaching errors. In (a) and (b), the magnitude of the forces increases with the grayscale, though equal grayscale do not correspond with equal force magnitude in both figures.

As explained in the introduction, apart from using the execution errors in subsequent movements to adapt to the visuomotor rotation, subjects could also use the muscular effort. For the HG group the "haptic channel" prevented the occurrence of execution errors. Subjects could push into the haptic wall, however as long as the force exerted on the end effector had a component in the direction of the target, the subject would reach the target. The force exerted in the direction perpendicular to the movement direction can be regarded as a waste of muscular effort. Minimizing the effort would be similar to minimizing these perpendicular interaction forces. We quantified the forces for the subjects in the HG as well as the SG and EE group by averaging them from the start to the end of every training movement during the training stage and the extended training stage. The start was defined as the moment when the velocity in the direction of the target last exceeded the 10 cm/s before reaching the maximum velocity and the end as the moment when the velocity first dropped below zero, after having reached the maximum velocity. Subsequently, the trial averages were averaged over the non-catch trials in subsequent cycles of five movements.

statistics

Baseline directional errors were compared by using an ANOVA with Group as between-subject factor. We tested whether the directional errors in the catch trails decreased in time and whether the groups differed in the amount of adaptation by using a repeated measures ANOVA with Group as between subject factor and Time (the repeated measure of the directional errors in the subsequent catch blocks) as within subject factor. A similar repeated measures ANOVA was used to assess differences in generalization by using the generalization catch blocks for the within subject factor Time. We performed a repeated measures ANOVA with group as between subject factor and the direction errors in catch block 6 and generalization catch block 1 as repeated measures to assess if the directional errors to the generalization targets deviated from the errors in the trained directions. Differences in after effects were assessed by using a ANOVA with Group as between-subject factor. For all significant main effects, post hoc tests (with Bonferroni adjustments for multiple comparisons) were performed to deduce which groups differed significantly from each other. To assess whether the interaction forces decreased significantly during training for the HG, SG and EE group, we performed a paired t-test for each separate group with the interaction forces in the first and last cycle as input. The level of significance was defined as $p<0.05$.

5.4 results

At the end of the familiarization stage, subjects of all groups were accustomed to the virtual environment and had learned to reach in the virtual environment (see Figure 5.4). Last movement cycles were used as baseline movements. Baseline trajectories were straight lines and the baseline directional error did not significantly differ between the different training groups ($p = 0.166$).

5.4.1 training trajectories

The effect of the applied forces on the hand path during early (first 25 movements) and late (last 25 movements) training is illustrated in Figure 5.4 for a representative subject of each group. The group A subjects did not experience any additional forces, so as expected these subjects showed hand paths that were initially directed roughly 30° counter clockwise to the target. During the course of the training the subjects adapted to the visuomotor rotation as evidenced by the approximately straight trajectories during late training. The error enhancing (group EE) and reducing (SG) forces led to larger and smaller curvatures of the initial hand paths, respectively. Also in these groups, the curvatures decreased during training. The aiming movements of group P and HG were forced along the optimal trajectory, resulting in absence of visible directional errors from the first training movement. The HG group had to generate the movement along the trajectory themselves, and consequently could show under and overshoot of the targets, as evidenced by hand paths passing the targets. On the contrary, the P group never showed under or overshoot, as these subjects were moved by the robot along the optimal trajectory.

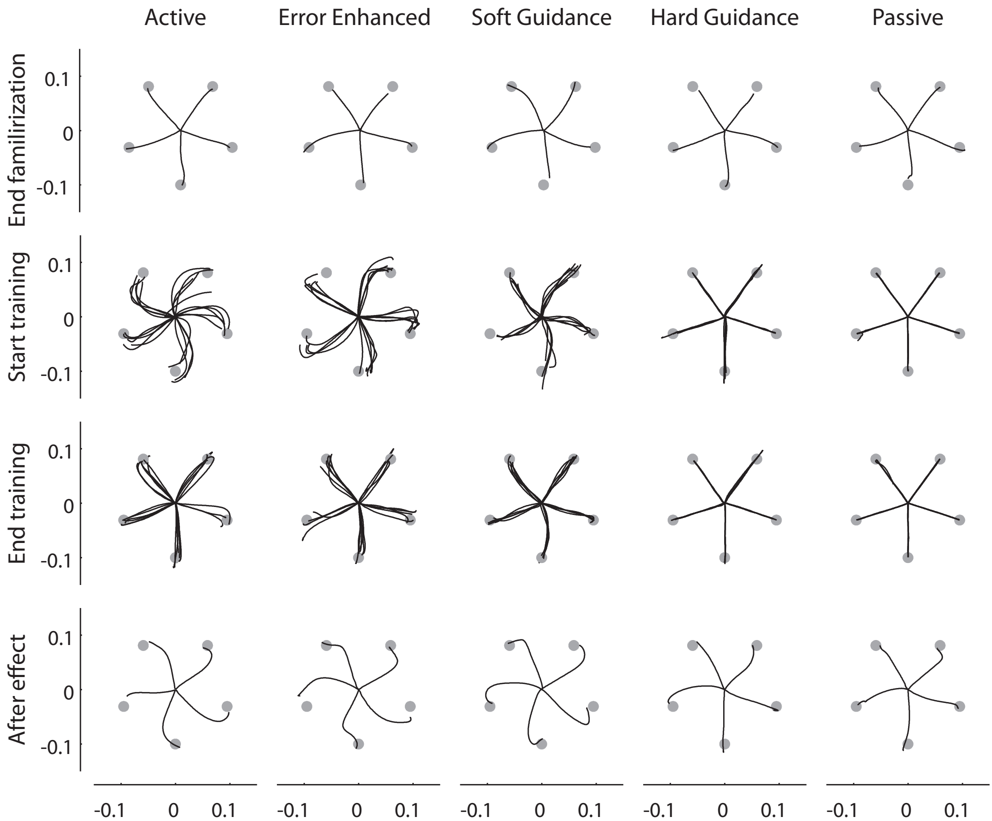

Figure 5.4. Representative hand-paths of a subject from each of the groups are compared to illustrate differences between hand paths during training (while they are exposed to the guiding forces) and wash out. The top row shows baseline hand paths, which are the last five movements during the familiarisation stage. Hand paths of the first 25 movements (with the exception of the catch trials) performed during the training stage are shown along the second row, whereas the hand paths of the last 25 movements of the training stage are shown along the third row. The bottom row shows the after effects, which occur during the first cycle of movements during the wash out phase.

5.4.2 **adaptation**

The presence of the force fields during the training movements made a direct comparison of the directional errors during these movements impossible. Therefore, directional errors made in the catch trials were used to assess the adaptation to the visuomotor rotation. Figure 5.5 shows the group averages for the different catch blocks. The repeated measures ANOVA showed that there was a significant main effect of Time ($p<0.001$) and Group ($p<0.001$) and a significant interaction effect of Group and Time ($p=0.015$) on the directional errors in the catch blocks. These results indicated that the groups on average adapted to the visuomotor rotation, that the directional errors in the different groups differed from each other and that the groups did not show similar changes over the different blocks of catch trials. Post hoc comparisons revealed that groups A and EE adapted the most to the visual rotation, expressed by directional errors

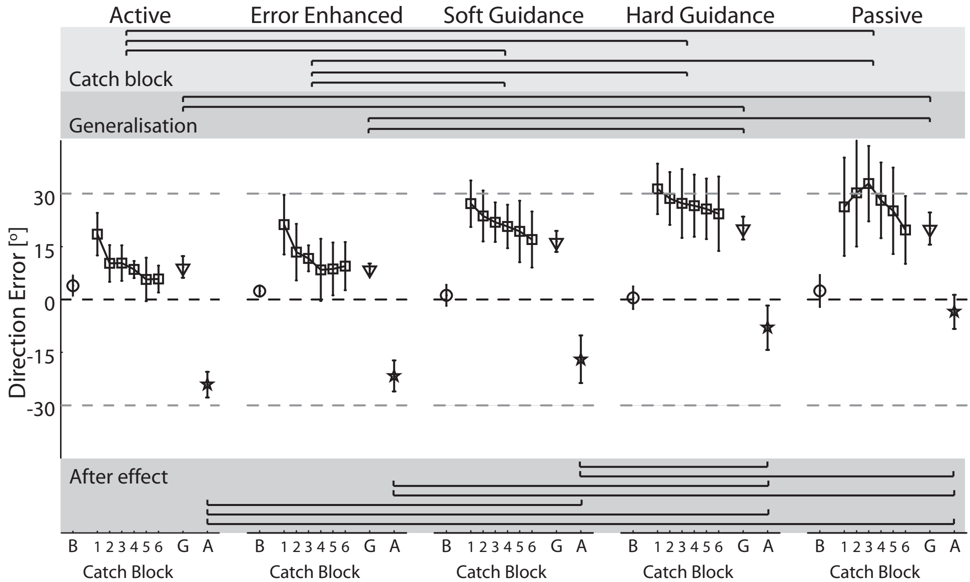

Figure 5.5. Mean directional errors for the different groups and for the different stages. The circles indicate the average directional errors over the last cycle of movement during the baseline. The squares show the average value of the subsequent blocks of catch trials during the training stage. For the generalisation catch trials, the average value over all blocks is depicted (triangle) as the different generalisation blocks did not differ significantly from each other. The stars show the average directional errors over the first cycle of movements during the wash out, the after effects. The error bars indicate the standard deviation. The horizontal brackets in the gray shading on the top and bottom indicate the significant differences (assessed with repeated measures ANOVA) between the groups in the overall average of the catch blocks, the generalisation and after effect.

that were significantly smaller than the errors of the SG ($p=0.002$ and $p=0.024$, respectively), HG ($p<0.001$ for A and EE) and P ($p<0.001$ for A and EE) group. The directional errors of the SG, HG, and P group did not differ significantly from each other.

In Figure 5.5, it can be seen that all groups, except P, show a gradual decrease of directional errors in the catch blocks, though the rate of change differed among groups, which was also expressed in the significant Group x Time interaction effect. To explore the different rates of adaptation further, we assessed for each combination of groups from which catch block onwards the groups showed a significant difference (see Table 5.1). Group A and EE did not differ significantly in the rate of adaptation as for none of the catch blocks a significant difference was found. The adaptation rate for these groups was higher as that for SG, HG, and P group. The SG group showed an intermediate adaptation rate as on one side it took longer before A and EE showed a significantly smaller directional error for SG than for the HG and P group, whereas on the other side the SG, HG and P group did not show any lasting significant difference.

In addition to the execution errors during the catch trials, adaptation could also be derived from the average forces on the hand during the training trials. As the exerted forces on the hand were dependent on the deviation from the optimal path, a decrease of the forces would be indicative for adaptation. Figure 5.6 shows the forces on the hand (averaged across subjects) as a function of the cycle. The average hand forces in the EE group were larger as those in the SG and HG groups; this could be mainly attributed to the unstable character of the EE force field, in which forces were directed to further increase execution errors. The difference in magnitude between the SG and HG group could be attributed to the larger stiffness used in the HG group. The gradual, significant ($p<0.001$) decrease of the forces in the SG and EE group confirmed the results of the catch trials that subjects in these groups clearly adapted to the visuomotor rotation. Remarkably, the HG group also showed a significant decrease ($p<0.001$) in the forces on the hand, which would point at the presence of adaptation. This adaptation is not likely to be driven by the execution errors, as the "haptic channel" in the HG prevented the occurrence of large execution errors. The average hand force in the first and last cycle were 6.5 and 2.7 N, which with the used stiffness of 5000 N/m is equivalent to deviations of the optimal path of 1.3 mm and 0.54 mm, respectively.

5.4.3 generalization

By using generalization catch trials we assessed how well subjects were able to use the learned visuomotor rotation in reaching to targets which were positioned exactly in between each pair of adjacent training targets. All groups performed equally well on reaching to the training targets as to the generalization targets, as there was neither a significant main effect of the repeated measure nor an interaction effect. Therefore, it can be expected that the differences between groups found in the generalization catch blocks are similar to those of the catch blocks. A second repeated measures ANOVA (with the six generalization catch blocks as repeated measure) confirmed these expectations. There was a main effect of Group ($p=0.001$) and post hoc tests showed that directional errors for Group A and EE, were significantly smaller than HG ($p=0.023$ and $p=0.014$) and P ($p=0.027$ and $p=0.016$) and that directional errors of SG, HG and P did not significantly differ from each other (see horizontal brackets in Figure 5.5). The only difference was, that A and EE did not differ significantly from SG during generalization. Apart of these group effects, there was no main effect of Time or a Group x Time interaction effect. The absence of these effects indicated that none of the groups significantly increased their ability to reach to the alternative directions during the extended training stage.

Table 5.1. Rate of adaptation. The table entries indicate the of number of the catch block from which on the directional error of the group indicated on the top was significantly ($p<0.05$) smaller than the group indicated on the left side. A smaller directional error is an indication for a faster adaptation to the visuomotor rotation. A "–" indicates that there was no statistical significance between these groups for any of the catch blocks.

	A	EE	SG	HG
P	C2	C2	-	-
HG	C1	C2	-	
SG	C2	C3		
EE	-			

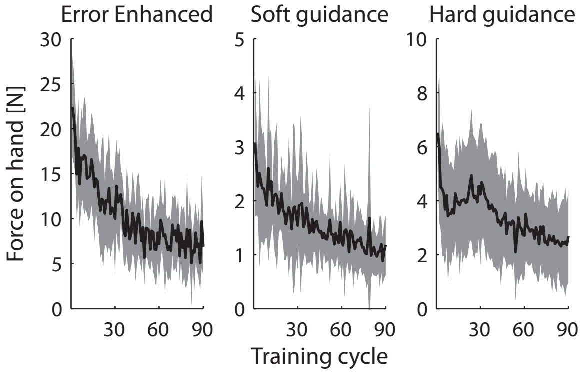

Figure 5.6. Average forces on the hand during the different training cycles for subjects in the Error Enhanced, Soft Guidance and Hard Guidance group. Each point on the curve represents the mean of subjects means, across the five movements within a cycle (excluding the catch trails and generalisation trials). The shading around the solid line indicates the standard deviation across the subjects. The first 60 training cycles are part of the training stage, whereas the last 30 cycles are part of the extended training stage.

5.4.4 **after effects**

The amount of adaptation in the preceding training stages was assessed by determining the after effects for the different groups when the visual rotation and any of the present force fields were turned off during the washout stage. The bottom row of graphs of Figure 5.4 shows the hand paths of the first five movements for a representative subject of each group. Group A and EE showed clear after affects as their hand paths could be regarded as mirrored trajectories of those shown during initial training, as if they were learning a 30° clockwise rotation. The hand paths also show a late "hook" back towards the end of the motion, which are likely the result of feedback mechanisms. For the SG group, the hand paths also showed a clear after effect and a "hook" back, though the effects were smaller. The hand paths of the HG and P group also showed slight clockwise curvature in some of the reaching directions.

The after effects were quantified by averaging the directional errors during the first five reaching movements in the wash out stage (see Figure 5.5 for group averages). An after effect occurred when the directional errors during initial washout differed from the directional errors during baseline. We performed a repeated measures ANOVA with Group as between subject factor and baseline and average washout score as repeated measures to assess which groups showed an after effect. The test showed a significant Time ($p<0.001$) and Time x Group ($p<0.001$) effect, indicating that directional errors during washout were significantly different from directional errors during baseline and that this difference was not equally large for all groups. Yet, post hoc comparisons turned out that for all groups this difference was significant (P: $p=0.001$, other groups: $p<0.001$), and as a consequence all groups showed after effects.

To compare the magnitude of the after effects between groups, a one way ANOVA was conducted. The ANOVA showed that there was a main effect of Group ($p<0.001$). The post hoc tests mainly confirmed the previously described differences in adaptation between the groups based on the catch trials. The effect between Group A, EE, P and HG were similar: Group A and EE (p=1.00) and Group P and HG (p=0.643) did not differ significantly from each other, whereas group A and group EE showed significantly larger after effects than group P (p<0.001 for A and EE) and HG (p<0.001 for A and EE). The main deviation with the adaptation results from the catch trials lies in the comparisons between the SG group and the other groups. The intermediate status of the SG group between Groups A and EE on one side and Groups P and HG was now also supported by significant differences and not merely from the absence of differences with groups on either of the sides. The after effect of SG was significantly larger than the after effect of the HG (p=0.005) and P(p<0.001) Group and was smaller compared to the A group (p=0.039) and EE group (p=0.528), though the last effect was not significant.

5.5 discussion

In this study, we investigated the effect of providing mechanical guidance during adaptation to a visuomotor rotation. The amount and direction of the provided guidance was manipulated through the use of force fields that differed in their dependency on the magnitude of the execution errors. Our data seem to provide support for the hypothesis that the provided interaction forces can be used in adaptation. However, they are not used as effectively as the execution errors, as they cannot prevent a decrease in the rate of adaptation as a consequence of reduced execution errors while receiving guidance. In the next paragraphs, we will further examine the role of execution errors and minimization of interactions forces/muscular effort.

5.5.1 execution errors

The magnitude of the execution errors seems to be the dominant driving factor in visuomotor adaptation. Reduction of the execution error, as was done in the SG group, resulted in a decreased amount and rate of adaptation and generalization compared to the A group. For the HG group the execution errors were further restrained to practically zero. This resulted in an absence of a significant adaptation, though the directional errors during the catch trials showed a slight gradual decrease. As the HG showed a small after effect, this slight decrease might have been an expression of adaptation occurring at a very slow rate. As execution errors during training could not explain this adaptation, a different error signal was at the origin of this adaptation. This will be discussed in more detail later on.

A decrease of execution errors resulted in smaller adaptation. However, when the execution errors were enlarged by using error enhancing forces, the larger execution errors did not seem to lead to a faster or larger adaptation. There are two possible explanations for this absence of increased adaptation. First, our assessment of adaptation might not have been sensitive for subtle changes in the rate of adaptation. We used catch trials every tenth movement to monitor adaptation and averaged the directional error of five subsequent catch trials. Consequently, the first data point during exposure reflected the average performance over the first 50 movements (=10 cycles). Previous studies (Krakauer *et al.*, 2000; Caithness *et al.*, 2004) have shown that during the first 10 cycles adaptation occurs at its highest rate. To have a closer look at these cycles we omitted the averaging and compared the directional errors in the first 5 catch trials between the A and EE group by using a repeated measures ANOVA. This analysis showed that the directional errors did not differ between the groups on any of the catch trials, further confirming that adaptation occurred at an equal rate for these groups.

Second, the error-enhanced forces might have changed the nature of the task. As the forces increased the reaching error, they made the task inherent unstable. Burdet and colleagues (Burdet *et al.*, 2001; Franklin *et al.*, 2003) demonstrated that executing arm movements in an unstable dynamic environment resulted in an increase of the impedance in the direction of the instability. Therefore, subjects that attended the EE training program might have adapted their impedance during reaching, in addition to the adaptations in the pointing direction. The adaptations in impedance might have slowed down the adaptation to the visuomotor rotation per se.

Previous studies (Emken & Reinkensmeyer, 2005; Wei *et al.*, 2005) have shown a beneficial effect of error augmentation. In stead of using forces to augment the error, Wei and colleagues (2005) implemented error augmentation during learning of a visuomotor rotation by providing visual feedback in which the deviations from the optimal straight trajectory were amplified with a gain of 2. They showed that a group of subjects receiving visual error augmentation during learning had a more than twice as large learning rate than a control group. So, a pure magnification of execution errors without "disturbing forces" does have a positive effect on adaptation.

5.5.2 **muscular effort**

The setup of this experiment was unique in that force fields were used to manipulate the execution errors. The forces were dependent on the trajectory errors. This is fundamentally different from the forces used in studies assessing adaptations to altered dynamics, as these forces are dependent on the position, velocity or acceleration of the hand. Recently, different studies have assessed whether learning to compensate for applied forces interferes with learning of a visuomotor task (Flanagan *et al.*, 1999; Krakauer *et al.*, 1999; Tong *et al.*, 2002). Although these studies showed that interference can occur if the applied forces and the kinematic perturbation depend on the same kinematic variable, i.e. hand position (Tong *et al.*, 2002), we believe that the forces in our study do not interfere with learning of the visuomotor rotation in this respect. Subjects did not need to compensate for the forces, as the forces decreased together with the adaptation to the rotation.

Instead of causing interference, we postulated that the interaction forces could have been used in adaptation to the visuomotor task. For the HG, SG and EE group minimizing the muscular effort by minimizing the interaction forces would have resulted in adaptation. The role of the interaction forces in driving adaptation can be best derived from the results of the HG group. During HG training the "haptic channel" prevented the occurrence of execution errors, by exerting forces on the hand. These interaction forces showed a small but gradual decrease during the course of training, showing that subjects little by little adapted their reaching direction, pushed less into the haptic wall and decreased the muscular effort of the movement. However, this adaptation occurred at a much slower rate than adaptation based on execution errors, as the adaptation was only clearly shown in the after effect and not in the catch blocks.

The results for the SG group also provided an indication for the minor role for optimizing dynamic criteria in adaptation. The directional errors during the catch trials seem to level off and not to continue decreasing to the magnitudes of the A and EE group. This lack of further improvement might be explained from the lack of the major driving stimulus during training. Indeed, when we further inspect the directional errors during the training trials, we see that the directional errors during the last cycle of the training stage are close to those of the A

Group (7.8° ± 2.4° vs. 3.5° ± 3.8°). So, at the end of the training, the guiding forces constrained the execution errors to levels almost equal to the end level of the A group. Although some more improvement based on the execution errors was possible, these errors will normally not return to baseline levels (Ghilardi *et al.*, 1995; Krakauer *et al.*, 2000; Klassen *et al.*, 2005). The SG group could have improved on the catch trials to a level equal to the A group when they would have decreased their reliance on the guiding forces. The guiding forces did continue to decrease during the extended training stage (see Figure 5.6). Still, the decrease occurred at a too slow rate, to bring the SG group to the level of the A group.

This was the first study to show that optimization of dynamic criteria can also contribute to adaptation to a visuomotor rotation. Dynamic criteria, like muscular effort or force have previously been shown to play a role in motor adaptation to novel dynamic environments (Scheidt *et al.*, 2000; Emken *et al.*, 2007b). Their results showed that minimizing effort is not as efficient as minimizing kinematic error in the process of adaptation. Our results extended their results to adaptation to a visuomotor task.

The minor role for force information in updating internal models was also shown by Milner and colleagues (2006). In their elegantly designed experiment, subjects first adapted to a position dependent force field with a parabolic profile. Subsequently subjects were exposed to a velocity-depend force field. This viscous force field required compensating forces in the opposite direction to the forces required for compensating the position dependent force field. Still, subject continued for about 15 trials to produce the compensating forces for the position dependent field, which effectively assisted the viscous force field. Based on these results, the authors concluded that information about the direction of the force seems not to be used in modifying the feed forward motor commands. However, possibly the dominant role of kinematic error in adaptation might have obscured to slow adaptation based on force information. In any case, Milner and Hinder (2006) argued that the marginal role of force information is due to the incapability of the peripheral force sensors to unambiguously indicate the direction of force.

5.5.3 **possible other mechanisms underlying adaptation**

Neither minimizing of the execution errors nor minimizing of the muscular effort can explain the small amount of adaptation in the P group, which was solely expressed in a small after effect. This implies that another mechanism underlies this adaptation. Candidate mechanisms are based on resolving a conflict between or within the different sensory modalities.

The rotation of the visual feedback resulted in a conflict between the proprioceptively and visually perceived hand location. This mismatch could be used to drive adaptation (Baraduc *et al.*, 2001). However, a major role of this mechanism is not likely as the magnitude of the inter-sensory discrepancy was equal for all groups (including the P group), whereas the groups showed different rates and levels of adaptation. Previous studies also provided support for the notion that this mechanism could at best play a minor role. Jones and colleagues (Jones *et al.*, 2001) showed that the muscle spindle activity was suppressed during visuomotor adaptation and therefore could not be used optimally. In addition, degrading of proprioceptive information by applying vibration little affected the adaptation to a visual distortion (Pipereit *et al.*, 2006). Nevertheless, even when this inter-sensory discrepancy only plays a minor it could explain the small after effects for the P as well as the HG group.

Another possible driving force of adaptation is the mismatch between the proprioceptive feedback and the expected proprioceptive feedback, which is calculated based on the efference copy. This mismatch occurs when the force fields considerably influence the intended movement direction, as is the case for the HG group. However also for the P group, this mismatch can occur. When the subjects in this group actively plan their movement on appearance of the targets, an efference copy can be generated and the accompanying expected proprioception can be calculated. The mismatch between actual and expected proprioceptive information is generally believed to underlie adaptation to a new dynamic force field (Tong *et al.*, 2002; Pipereit *et al.*, 2006). In that case motor commands are adjusted such that the actual proprioception will gradually change towards the expected proprioception of a straight movement. However, in our study, this process would not be efficient, as the enforced actual proprioceptive feedback is the desired feedback. And consequently adaptation based on getting the actual feedback towards the expected feedback would reinforce moving in the baseline directions.

Another possibility is that subjects in the P group used a cognitive strategy during training. This could explain the large variation in the response in the catch trials of the subjects in the P as some of these subjects might have been aware of the rotation and used a cognitive strategy during catch trials, whereas others did not. The use of cognitive strategies can explain the large variation, however it is not likely that it can explain the small after effect. Recently, Mazzoni and colleagues (2006) studied the use of an explicit strategy in learning to adapt to a visuomotor rotation. The cognitive strategy resulted in an immediate increase of the performance. However, over time the directional errors increased again, which could be regarded as evidence for the implicit learning processes overriding the explicit strategy.

In short, the inter-sensory discrepancy could have been responsible for the slow adaptation in the P and HG group, whereas it is not likely that the proprioceptive intra-sensory discrepancy resulted in any adaptation. Although in the previous section we argued that the decrease of the forces in the HG group could be regarded as evidence in support for the minimization of effort in adaptation. This decrease could in fact also be a secondary effect of resolving the inter-sensory mismatch. Based on the results of this study we cannot discern whether minimizing the muscular effort and/or the inter-sensory discrepancy drives the small adaptation seen in HG.

5.5.4 **generalization**

The generalization catch trials showed that the direction error during reaching to untrained directions were not statistically significant from reaching to trained directions. So by only learning five directions, the subjects were able to interpolate the locally learned directions, to the intermediate directions without significant degradation of the performance. The difference in performance on the generalization catch trials between the groups could be explained by the differences in performance on the catch trials. Therefore, it can be concluded that the difference between the groups in nature and extent of the error signals, only affected generalization through the amount of adaptation and did not affect generalization in a different way for example by inducing more locally learned directions.

5.5.5 implications for motor (re)learning in neurological rehabilitation

Robotic devices are increasingly popular to provide guiding forces, similar to the ones used in this study, to support the impaired movements during training of stroke patients (Hogan & Krebs, 2004). The use of robotic devices has been promoted from the notion that relearning the control of movements in stroke patients is akin to motor learning (Hogan *et al.*, 2006; Krakauer, 2006). Systematic overviews of clinical effect studies have shown that the effect of training in stroke patients, like motor learning is task specific and is largely depended on the intensity(Kwakkel *et al.*, 1999; Van Peppen *et al.*, 2004). These results are in concordance with studies showing that possibly similar neural correlates (Ward, 2006) underlie recovery (Ward *et al.*, 2003)and motor learning (Hikosaka *et al.*, 2002). The exact nature of neural plasticity is not yet known, yet it seems that repetitive time correlated motor and sensory stimulation of brain regions is required.

Different algorithms have been implemented to calculate the guiding forces to facilitate movements during training, including algorithms similar to “soft guidance” (Aisen *et al.*, 1997), “hard guidance” (Kahn *et al.*, 2006) and “passive” (Hesse *et al.*, 2003; Lynch *et al.*, 2005). Based on the results of our study, we would suggest that for optimal relearning, patients can best initiate and generate the movements themselves and in doing so are free to make execution errors. Still, guiding forces should be used to keep the execution errors of growing too large. Furthermore, the results of the SG group in our study showed that subjects relied on the guiding forces in restraining the execution errors which slowed down further adaptation. Therefore, to prevent reliance on guiding forces, we suggest that the amount of support should be progressively lowered in the course of rehabilitation when patient increase their performance.

Hitherto, most effect studies of robot-aided training concentrated on comparing robotic therapy to conventional therapies (Volpe *et al.*, 1999; Lum *et al.*, 2002; Prange *et al.*, 2006) and although the different algorithms mostly showed favorable effects, none of them showed superior effects over the others. However, one study did (Ferraro *et al.*, 2003) and this study showed results in agreement with our suggestions. In this study, an algorithm was used that adapted the amount of assistive forces during the course of rehabilitation to the motor abilities of the patients. Patients trained with this performance-based progressive therapy showed larger decreases of impairments compared to stroke patients whose assistive forces were not adapted (Hogan *et al.*, 2006).

5.5.6 conclusions

We conclude that applying guidance does not have a positive effect on adaptation to a visuomotor rotation. When guiding resulted in a decrease of the execution errors, the applied assistance cannot substitute for this decrease in driving adaptation. Restraining of the execution errors during either active of passive movements showed that minimization of muscular effort or the mismatch between visual and proprioceptive feedback could also be responsible for adaptation, however at a much lower rate. The less efficient use of muscular effort compared to execution errors, is in accordance with the minor role of minimizing muscular effort in adaptation to a new dynamic environment.

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5.7 references

Aisen ML, Krebs HI, Hogan N, McDowell F & Volpe BT. (1997). The effect of robot-assisted therapy and rehabilitative training on motor recovery following stroke. *Archives of Neurology* 54: 443-446.

Baraduc P, Guigon E & Burnod Y. (2001). Recoding arm position to learn visuomotor transformations. *Cerebral Cortex* 11: 906-917

Burdet E, Osu R, Franklin DW, Milner TE & Kawato M. (2001). The central nervous system stabilizes unstable dynamics by learning optimal impedance. *Nature* 414: 446-449

Caithness G, Osu R, Bays P, Chase H, Klassen J, Kawato M, Wolpert DM & Flanagan JR. (2004). Failure to consolidate the consolidation theory of learning for sensorimotor adaptation tasks. *The Journal of Neuroscience* 24: 8662-8671

Colombo G, Joerg M, Schreier R & Dietz V. (2000). Treadmill training of paraplegic patients using a robotic orthosis. *Journal of Rehabilitation Research and Development* 37: 693-700.

Donchin O, Francis JT & Shadmehr R. (2003). Quantifying generalization from trial-by-trial behavior of adaptive systems that learn with basis functions: theory and experiments in human motor control. *The Journal of Neuroscience* 23: 9032-9045

Emken JL, Benitez R & Reinkensmeyer DJ. (2007a). Human-robot cooperative movement training: Learning a novel sensory motor transformation during walking with robotic assistance-as-needed. *Journal of Neuroengineering and Rehabilitation* 4: 8

Emken JL, Benitez R, Sideris A, Bobrow JE & Reinkensmeyer DJ. (2007b). Motor Adaptation as a Greedy Optimization of Error and Effort. *Journal of Neurophysiology*

Emken JL & Reinkensmeyer DJ. (2005). Robot-enhanced motor learning: accelerating internal model formation during locomotion by transient dynamic amplification. *IEEE Transactions on Neural Systems and Rehabilitation Engineering* 13: 33

Ferraro M, Palazzolo JJ, Krol J, Krebs HI, Hogan N & Volpe BT. (2003). Robot-aided sensorimotor arm training improves outcome in patients with chronic stroke. *Neurology* 61: 1604-1607

Flanagan JR, Nakano E, Imamizu H, Osu R, Yoshioka T & Kawato M. (1999). Composition and decomposition of internal models in motor learning under altered kinematic and dynamic environments. *The Journal of Neuroscience* 19: RC34.

Franklin DW, Osu R, Burdet E, Kawato M & Milner TE. (2003). Adaptation to stable and unstable dynamics achieved by combined impedance control and inverse dynamics model. *Journal of Neurophysiology* 90: 3270-3282

Ghez C, Krakauer JW, Sainburg RL & Ghilardi M-F. (2000). Spatial representation and internal models of limb dynamics in motor learning. In *The new cognitive neurosiences*, ed. Gazzaniga MS, pp. 501-514. MIT press, Cambridge.

Ghilardi MF, Gordon J & Ghez C. (1995). Learning a visuomotor transformation in a local area of work space produces directional biases in other areas. *Journal of Neurophysiology* 73: 2535-2539

Goodbody SJ & Wolpert DM. (1998). Temporal and amplitude generalization in motor learning. *Journal of Neurophysiology* 79: 1825-1838.

Hesse S, Schulte-Tigges G, Konrad M, Bardeleben A & Werner C. (2003). Robot-assisted arm trainer for the passive and active practice of bilateral forearm and wrist movements in hemiparetic subjects. *Archives of Physical Medicine and Rehabilitation* 84: 915-920

Hikosaka O, Nakamura K, Sakai K & Nakahara H. (2002). Central mechanisms of motor skill learning. *Current Opinion in Neurobiology* 12: 217-222

Hogan N & Krebs HI. (2004). Interactive robots for neuro-rehabilitation. *Restorative Neurology and Neuroscience* 22: 349-358

Hogan N, Krebs HI, Rohrer B, Palazzolo JJ, Dipietro L, Fasoli SE, Stein J, Hughes R, Frontera WR, Lynch D & Volpe BT. (2006). Motions or muscles? Some behavioral factors underlying robotic assistance of motor recovery. *Journal of Rehabilitation Research and Development* 43: 605-618

Jones KE, Wessberg C & Vallbo A. (2001). Proprioceptive feedback is reduced during adaptation to a visuomotor transformation: preliminary findings. *Neuroreport* 12: 4029-4033

Kahn LE, Lum PS, Rymer WZ & Reinkensmeyer DJ. (2006). Robot-assisted movement training for the stroke-impaired arm: Does it matter what the robot does? *Journal of Rehabilitation Research and Development* 43: 619-630

Klassen J, Tong C & Flanagan JR. (2005). Learning and recall of incremental kinematic and dynamic sensorimotor transformations. *Experimental Brain Research* 164: 250-259

Krakauer JW. (2006). Motor learning: its relevance to stroke recovery and neurorehabilitation. *Current Opinion in Neurology* 19: 84-90

Krakauer JW, Ghilardi MF & Ghez C. (1999). Independent learning of internal models for kinematic and dynamic control of reaching. *Nature Neuroscience* 2: 1026-1031

Krakauer JW, Pine ZM, Ghilardi MF & Ghez C. (2000). Learning of visuomotor transformations for vectorial planning of reaching trajectories. *The Journal of Neuroscience* 20: 8916-8924

Kwakkel G, van Peppen R, Wagenaar RC, Wood Dauphinee S, Richards C, Ashburn A, Miller K, Lincoln N, Partridge C, Wellwood I & Langhorne P. (2004). Effects of augmented exercise therapy time after stroke: a meta-analysis. *Stroke* 35: 2529-2539

Kwakkel G, Wagenaar RC, Twisk JW, Lankhorst GJ & Koetsier JC. (1999). Intensity of leg and arm training after primary middle-cerebral-artery stroke: a randomised trial. *Lancet* 354: 191-196.

Lum PS, Burgar CG, Shor PC, Majmundar M & Van der Loos M. (2002). Robot-assisted movement training compared with conventional therapy techniques for the rehabilitation of upper-limb motor function after stroke. *Archives of Physical Medicine and Rehabilitation* 83: 952-959

Lynch D, Ferraro M, Krol J, Trudell CM, Christos P & Volpe BT. (2005). Continuous passive motion improves shoulder joint integrity following stroke. *Clinical Rehabilitation* 19: 594-599

Mazzoni P & Krakauer JW. (2006). An implicit plan overrides an explicit strategy during visuomotor adaptation. *The Journal of Neuroscience* 26: 3642-3645

Milner TE & Hinder MR. (2006). Position information but not force information is used in adapting to changes in environmental dynamics. *Journal of Neurophysiology* 96: 526-534

Pipereit K, Bock O & Vercher JL. (2006). The contribution of proprioceptive feedback to sensorimotor adaptation. *Experimental Brain Research* 174: 45-52

Prange GB, Jannink MJ, Groothuis-Oudshoorn CG, Hermens HJ & Ijzerman MJ. (2006). Systematic review of the effect of robot-aided therapy on recovery of the hemiparetic arm after stroke. *Journal of Rehabilitation Research and Development* 43: 171-184

Sainburg RL & Wang J. (2002). Interlimb transfer of visuomotor rotations: independence of direction and final position information. *Experimental Brain Research* 145: 437-447.

Scheidt RA, Conditt MA, Secco EL & Mussa-Ivaldi FA. (2005). Interaction of visual and proprioceptive feedback during adaptation of human reaching movements. *Journal of Neurophysiology* 93: 3200-3213

Scheidt RA, Dingwell JB & Mussa-Ivaldi FA. (2001). Learning to move amid uncertainty. *Journal of Neurophysiology* 86: 971-985

Scheidt RA, Reinkensmeyer DJ, Conditt MA, Rymer WZ & Mussa-Ivaldi FA. (2000). Persistence of motor adaptation during constrained, multi-joint, arm movements. *Journal of Neurophysiology* 84: 853-862

Shadmehr R & Mussa-Ivaldi FA. (1994). Adaptive representation of dynamics during learning of a motor task. *The Journal of Neuroscience* 14: 3208-3224

Teasell R, Bitensky J, Salter K & Bayona NA. (2005). The role of timing and intensity of rehabilitation therapies. *Topics in Stroke Rehabilitation* 12: 46-57

Thoroughman KA & Shadmehr R. (1999). Electromyographic correlates of learning an internal model of reaching movements. *The Journal of Neuroscience* 19: 8573-8588.

Thoroughman KA & Shadmehr R. (2000). Learning of action through adaptive combination of motor primitives. *Nature* 407: 742-747.

Tong C, Wolpert DM & Flanagan JR. (2002). Kinematics and dynamics are not represented independently in motor working memory: evidence from an interference study. *The Journal of Neuroscience* 22: 1108-1113.

Van Peppen RP, Kwakkel G, Wood-Dauphinee S, Hendriks HJ, Van der Wees PJ & Dekker J. (2004). The impact of physical therapy on functional outcomes after stroke: what's the evidence? *Clinical Rehabilitation* 18: 833-862

Volpe BT, Krebs HI, Hogan N, Edelsteinn L, Diels CM & Aisen ML. (1999). Robot training enhanced motor outcome in patients with stroke maintained over 3 years. *Neurology* 53: 1874-1876

Wang J & Sainburg RL. (2005). Adaptation to visuomotor rotations remaps movement vectors, not final positions. *The Journal of Neuroscience* 25: 4024-4030

Ward NS. (2006). The neural substrates of motor recovery after focal damage to the central nervous system. *Archives of Physical Medicine and Rehabilitation* 87: S30-35

Ward NS, Brown MM, Thompson AJ & Frackowiak RS. (2003). Neural correlates of motor recovery after stroke: a longitudinal fMRI study. *Brain* 126: 2476-2496

Wei Y, Bajaj P, Scheidt R & Patton J. (2005). Visual error augmentation for enhancing motor learning and rehabilitative relearning. In *Proceedings of ICORR 2005 - IEEE International Conference on Rehabilitation Robotics*: 505-510. Chicago, Illinois, USA.

Wu G, van der Helm FC, Veeger HE, Makhsous M, Van Roy P, Anglin C, Nagels J, Karduna AR, McQuade K, Wang X, Werner FW & Buchholz B. (2005). ISB recommendation on definitions of joint coordinate systems of various joints for the reporting of human joint motion--Part II: shoulder, elbow, wrist and hand. *Journal of Biomechanics* 38: 981-992

chapter 6

the effects on kinematics and muscle activity of walking in a robotic gait trainer during zero-force control

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6.1 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 DoF 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.

6.2 **introduction**

Development of robotic devices for gait rehabilitation of stroke patients is motivated by the need for an intensive training, which has been shown to be the key element in facilitating recovery (Kwakkel *et al.*, 1997; Kwakkel *et al.*, 2004; Teasell *et al.*, 2005), together with the need for a therapist-friendly training. Now that several robotic devices for gait rehabilitation, i.e. Lokomat (Colombo *et al.*, 2000) and Gait Trainer (Hesse *et al.*, 1999), are available on the market (Krebs & Hogan, 2006), most research effort is put in determining the effectiveness of these devices (Husemann *et al.*, 2007; Pohl *et al.*, 2007) and in introducing new principles and concepts to extend the possibilities compared to these "first generation" devices (Riener *et al.*, 2005; Agrawal *et al.*, 2007; Aoyagi *et al.*, 2007; Schmidt *et al.*, 2007b; Von Zitzewitz *et al.*, 2007). These "first generation" devices are generally 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 (Husemann *et al.*, 2007; Pohl *et al.*, 2007) , treatment outcome could be further optimized by increasing the active participation of the patient.

Active patient participation may be realized by increasing motivation (Lunenburger *et al.*, 2007) 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 and colleagues (Cai *et al.*, 2006), 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 previously used strategy of active assistance(Ferraro *et al.*, 2003; Krebs *et al.*, 2003; Hogan *et al.*, 2006).

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 some kind of interaction control, usually called haptic- or impedance-control(Aoyagi *et al.*, 2007). By adjusting the impedance, the behaviour of an impedance controlled robot can be varied from very stiff to very flexible. A very stiff control mode would then resemble a position control as implemented in existing devices. 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 AAN control or selective control (Van Asseldonk *et al.*, 2007) algorithm. Only assisting 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 around that joint to overcome this resistance. However when walking without the device the same 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 (Dobkin, 2004; Van Peppen *et al.*, 2004; Bayona *et al.*, 2005), 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, although optionally restrained by control of the robot. Consequently, the evaluation of a patient in charge mode reveals at which points the device still influences unrestrained walking. 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 (Hidler *et al.*, 2005; Israel *et al.*, 2006; Neckel *et al.*, 2006; Neckel *et al.*, 2007). The Lokomat can also function in a low impedance mode, but only preliminary results have been published (Riener *et al.*, 2005) . 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 (Van der kooij *et al.*, 2006; Veneman *et al.*, 2007a). 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 (Israel *et al.*, 2006). The importance of adding pelvic motions to gait training, has recently also been pointed out by Aoyagi and Colleagues (Aoyagi *et al.*, 2007) 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 AAN 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.

6.3 materials and methods

6.3.1 subjects

Ten healthy young adults (4 male, 6 female), mean age 25.9 years volunteered to be participants for this experiment. All participants provided informed consent before testing began.

6.3.2 experimental apparatus and recordings

rehabilitation device

For the experiments the prototype of the gait rehabilitation robot LOPES was used. LOPES is an exoskeleton type rehabilitation robot. It is lightweight and actuated by Bowden cable driven series elastic actuators (Veneman *et al.*, 2006). It offers 3D translations of the pelvis, where the antero-posterior and medio-lateral motions (1 and 2 in Figure 6.1a) are actuated. Furthermore, it has two actuated rotation axes in the hip joints (4 and 5 in Figure 6.1a), and actuated knees with one rotation axis (6 in Figure 6.1a) (Van der kooij *et al.*, 2006). 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 degree of freedom (Veneman *et al.*, 2006). The latter mode was used in this research to investigate to what degree "free" walking with the device resembled free walking on a treadmill. The pelvis and the legs of the subject are strapped to the exoskeleton (Figure 6.1b). The inertial characteristics of the robot are presented in Table 6.1. These are relevant because in the current version of the device we are not able to compensate for the inertia of the robotic limbs.

Table 6.1. Approximate inertial and mass characteristics of the robot frame. For comparison approximate values for a human person with a body mass of 75 kg. 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 kg	-	75 kg (45 kg trunk only)	-
	Sideways	27 kg	-	75 kg (45 kg trunk only)	-
Upper leg	Hip Flex/Ext	2.9 kg	0.088 kgm^2	9.5 kg	0.15 kgm^2
Lower leg	Knee Flex/Ext	2.2 kg	0.064 kgm^2	3.5 kg	0.06 kgm^2

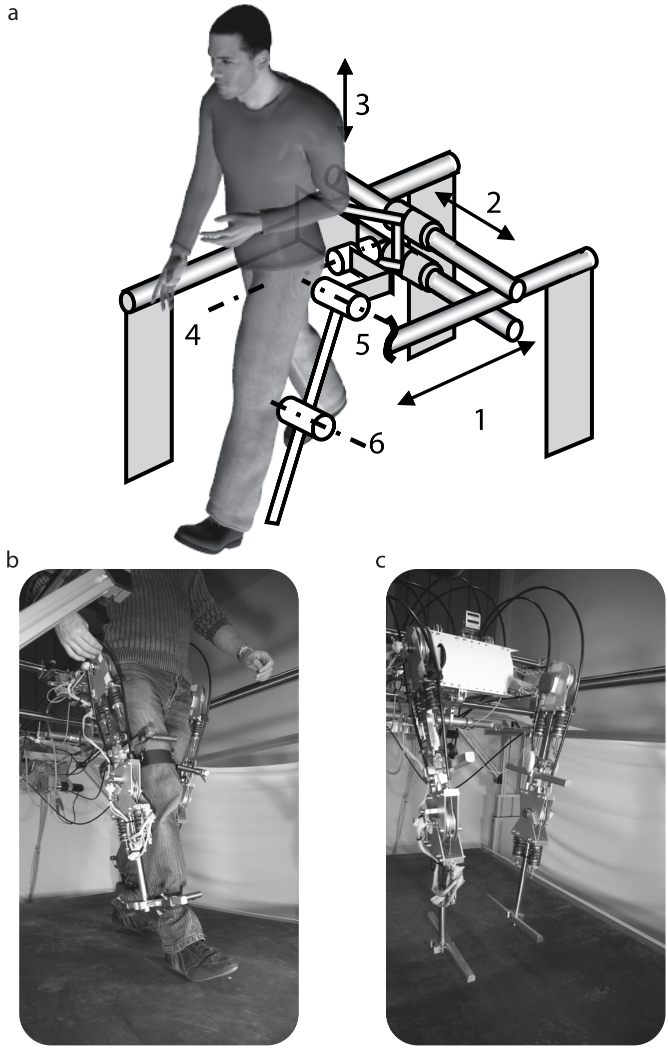

Figure 6.1. a) Schematic overview of the LOPES exoskeleton and its degrees of freedom. The horizontal movements of the pelvis (1 and 2) are actuated, as well as the movements of the leg joints (4, 5 and 6). The vertical movement of the pelvis is unactuated. The three degrees of the not depicted left exoskeleton leg complete the number of degrees of freedom to nine. b) Fixation of the subject in the LOPES exoskeleton . c) the exoskeleton with its bowden cables to transmit the forces from the motor to the joint.

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 and lower leg was measured by attaching frames with four infrared markers on the back of the thigh and shank. A frame of four markers was attached to the vertebra prominence, measuring the orientation and position of the trunk. 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 placing 4 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 by sending a pulse at the start and end of each measurement to both systems and aligning these pulses in time.

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 (shaving of hair, abrading, cleansing with alcohol) and the placement of the disc-shaped (35x26 mm) solid-gel Ag/AgCl-electrodes (type H93SG, Tyco Healthcare / Kendall, Mansfield, USA) in a bipolar configuration (interelectrode distance of 26 mm) were performed according to Seniam guidelines (Hermens *et al.*, 1999). For the EMG recordings a compact measurement apparatus (type Porti 16-5, TMS International, Enschede, The Netherlands) was used. The analog signals were not filtered before sampling at 1024 Hz, which was justified by using sigma-delta analog-to-digital converters with inherent anti aliasing filters. The signals were sent from the porTable 6.unit via fibre optics to the computer, where data were stored for further processing.

6.3.3 **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 m/s, 0.75 m/s and 1.25 m/s. The first two velocities were chosen to reflect often reported walking velocities of stroke patients and the latter velocity was chosen to reflect a normal walking velocity. 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 minutes 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. 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.

During this experiment 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. Due to the used control method (Veneman *et al.*, 2006) the torques caused by inertia and weight of the exoskeleton remain perceptible for the subject. However, the exoskeleton was designed to have low 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 (Figure 6.1B).

6.3.4 **data analysis**

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 (VZAnalyzer™ 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 we did not measure the pelvis angles, we could not calculate the hip angles. However, as the pelvis rotations are relatively small compared to the thigh rotations, 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 the angular range of motion of the frontal and sagittal trunk and thigh angle and the sagittal knee angle. The ranges were calculated by subtracting the minimum angle from the maximum angle for each step-cycle.

gait parameters

We also compared the spatial and temporal gait parameters (Perry, 1992; Hausdorff *et al.*, 1995) From the measured marker positions, we calculated the step width. 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.

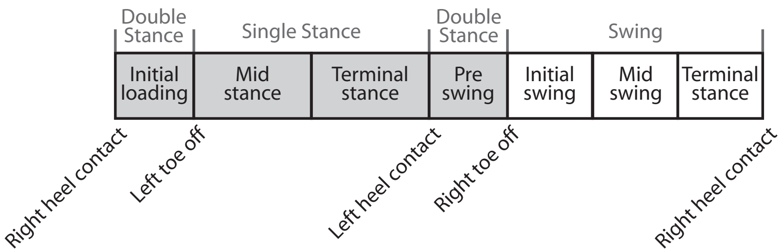

Figure 6.2. Segmentation of the gait cycle into phases for the calculation of the integrated EMG over these phases. For clarity, the standard gait events are indicated with respect to these phases.

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

EMG measurements

All EMG processing was done with custom written software (Roetenberg *et al.*, 2003) in Matlab (Nattick, USA). The raw EMG data were band pass filtered at 10-400 Hz with a 2nd order zero-lag Butterworth filter and converted to smooth rectified EMG signals (SRE) using a low-pass 2nd 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, based on subsequent heel strikes. 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 7 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 phases of walking were defined based on the footswitch data of both feet (see Figure 6.2). 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: mid stance and terminal stance and the swing phase was split up into 3 intervals of equal length: initial swing, mid swing and terminal swing.

The smoothed EMG data were used to determine the onset and cessation times of the main burst for each muscle. The used algorithm (Roetenberg *et al.*, 2003) is based on the approximated generalized likelihood ratio (AGLR) principle described by Staude and Wolf (Staude, 2001). 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.

6.3.5 statistical analysis

Linear mixed modelling 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, in the model 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 walking*velocity", "type of walking*interval" and "velocity*interval" and three-way interaction "type of walking*velocity*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.

6.4 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.

6.4.1 gait parameters and kinematics

The basic gait parameters showed some significant changes while walking in LOPES compared to treadmill walking (see Table 6.2). The cycle time and total stance time did not show a significant change, whereas the swing time showed a significant increase. The absence of a change in cycle time implied that also the step length did not change as these measures are directly related when walking with imposed velocities. The increase in swing time was accompanied by an increase in single stance time, which led to a significant decrease in Double stance ratio ($p<.001$). The step width showed a small (0.014 m) but significant ($p=.011$) increase in LOPES walking compared to treadmill walking.

The changes in gait parameters were accompanied by changes of the segment/joint kinematics. Figure 6.3 illustrates the ensemble average over all subjects for the rotations in trunk, thigh and knee. The sagittal knee and thigh rotations and the frontal thigh rotation showed similar patterns in LOPES walking and treadmill walking. Still, the range of motion for the knee

Table 6.2. Average values and within- subject variability of the gait parameters (± standard deviations) for the different types of walking

Variable	Average		Within-subject variability	
	Treadmill walking	LOPES walking	Treadmill walking	LOPES walking
Cycle time	1.427 ± .308 s	1.463 ± .326 s	.031 ± .020 s	.034 ± .023 s
Double stance ratio	.45 ± .05	.40 ± .06 *	.028 ± .017	.033 ± .020
Swing time	.472 ± .082 s	.508 ± .089 s *	.032 ± .015 s	.035 ± .027 s
Stance time	.955 ± .232 s	.955 ± .246 s	.018 ± .008 s	.018 ± .011 s
Step width	.328 ± .059 m	.342 ± .053 m *	.023 ± .006 m	.027 ± .010 m *

* significantly different from treadmill walking with $p<.05$

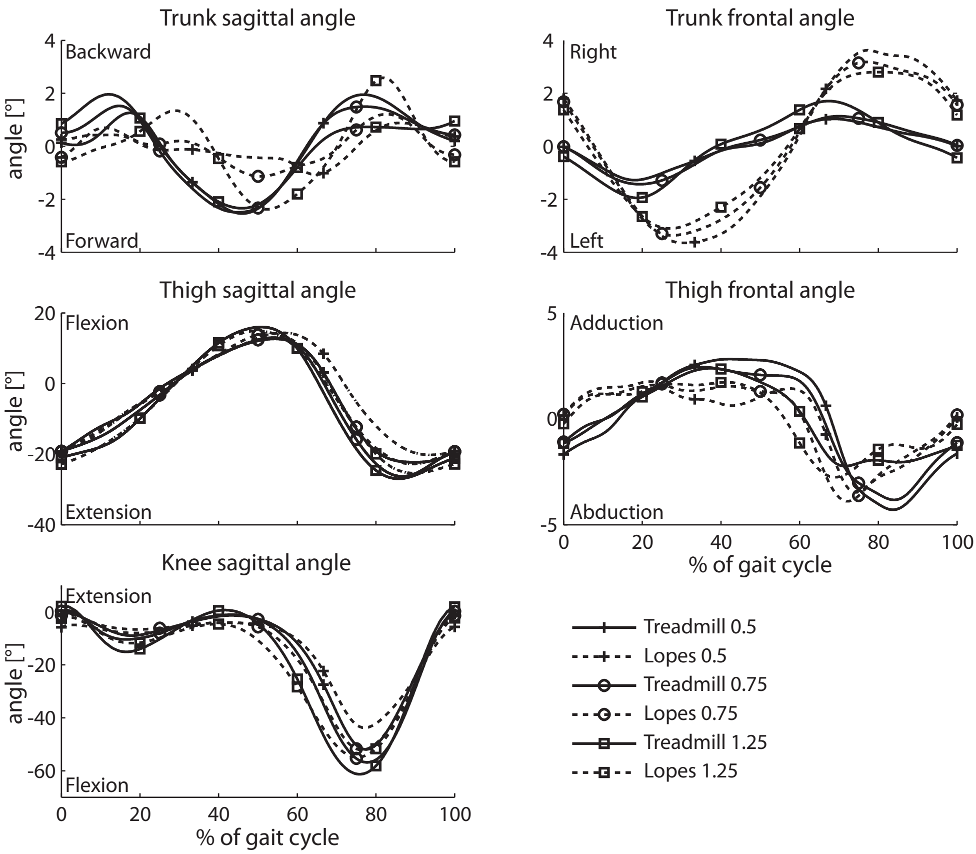

Figure 6.3. The average trajectories of the sagittal and frontal rotations of the trunk (upper panels), of the upper leg (middle panels) and the sagittal angle of the knee (lower panel) for all conditions, averaged over all subjects.

(p<.001) and sagittal (p=.005) thigh movements were significantly smaller in LOPES walking compared to treadmill walking (see Table 6.3). 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<.001) change of the range of motion. During LOPES walking subjects demonstrated an increased frontal trunk rotation.

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 Table 6.2 and 3). The variability in the step width (p=.032) and the sagittal trunk angle range (p=0.015) was significantly larger in LOPES walking compared to treadmill 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.

6.4.2 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.

onset and cessation times of burst

The onset and cessation times of the main bursts reflect the timing of muscle activity (see Figure 6.4). 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 (Nene *et al.*, 2004). The occurrence of the burst in rectus femoris was strongly dependent on the walking velocity. For the lowest velocity only one subject in each condition showed the burst (see numbers in the rectus femoris panel of Figure 6.4). The absence of a burst in the other subjects indicated that this muscle was not active for these subjects. Although these unbalanced numbers obstruct the performance of statistical analysis, the inequality in subjects showing the burst could be considered as a measure of the difference in treadmill and LOPES walking. Furthermore, these differences in burst occurrence will also be reflected in differences in mean integrated activity. 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 % vs. 53.3 % walking, $p<0.001$), whereas the cessation did not change significantly. Apart of the changes in rectus remoris, three significant changes were observed in the timing of the burst of the other muscles. The onset occurred significantly earlier in LOPES walking compared to treadmill walking for the vastus lateralis (88.0 % vs. 90.0 %, $p=0.031$). The burst of the gluteus medius ended significantly earlier in LOPES walking compared to treadmill walking (40.5 % vs. 45.4 %, $p<0.001$) whereas the biceps femoris showed a delay cessation while walking in LOPES (16.1 % vs 23.4 %, $p=0.024$). None of the muscles showed an interaction effect between velocity and condition.

Table 6.3. 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	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<.05

mean integrated activity

Although all muscles showed approximately the same pattern during LOPES walking and treadmill walking, (*this can be imagined by connecting the different bars with lines in Figure 6.5*) they all showed a significant difference in the integrated activity for at least one interval (an overview can be found in Table 6.5). 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 6.4). The difference was dependent on the interval. During the start and the end of the stance phase, the mean activity was significantly lower, while 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 the initial loading the activity of both muscles was lower in LOPES. However, using needle electrodes Nene and colleagues (2004) showed that the activity of the rectus femoris during initial loading and the single stance was the consequence of cross talk from the vastus lateralis and vastus intermedius. The significant 3rd order type of walking x interval x velocity 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.

Table 6.4. 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.

	Type of walking	Type of walking x Interval	Type of walking x Velocity	Type of walking x Interval x Velocity
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		

Table 6.5. The p-values for comparison of the mean normalized EMG of LOPES walking and treadmill walking for the different muscle and for the different intervals.

	Gl Ma	Gl me	RF	VL	TA	BF	AL	Ga Me
Initial loading	.004	<.001	<.001	<.001		<.001		.049
Mid stance						<.001	.012	
Terminal stance	.002	<.001					.002	<.001
Pre swing	.007		.012		.009		.002	
Initial swing			.017		<.001			
Mid swing						.002		
Terminal swing					.005	<.001		

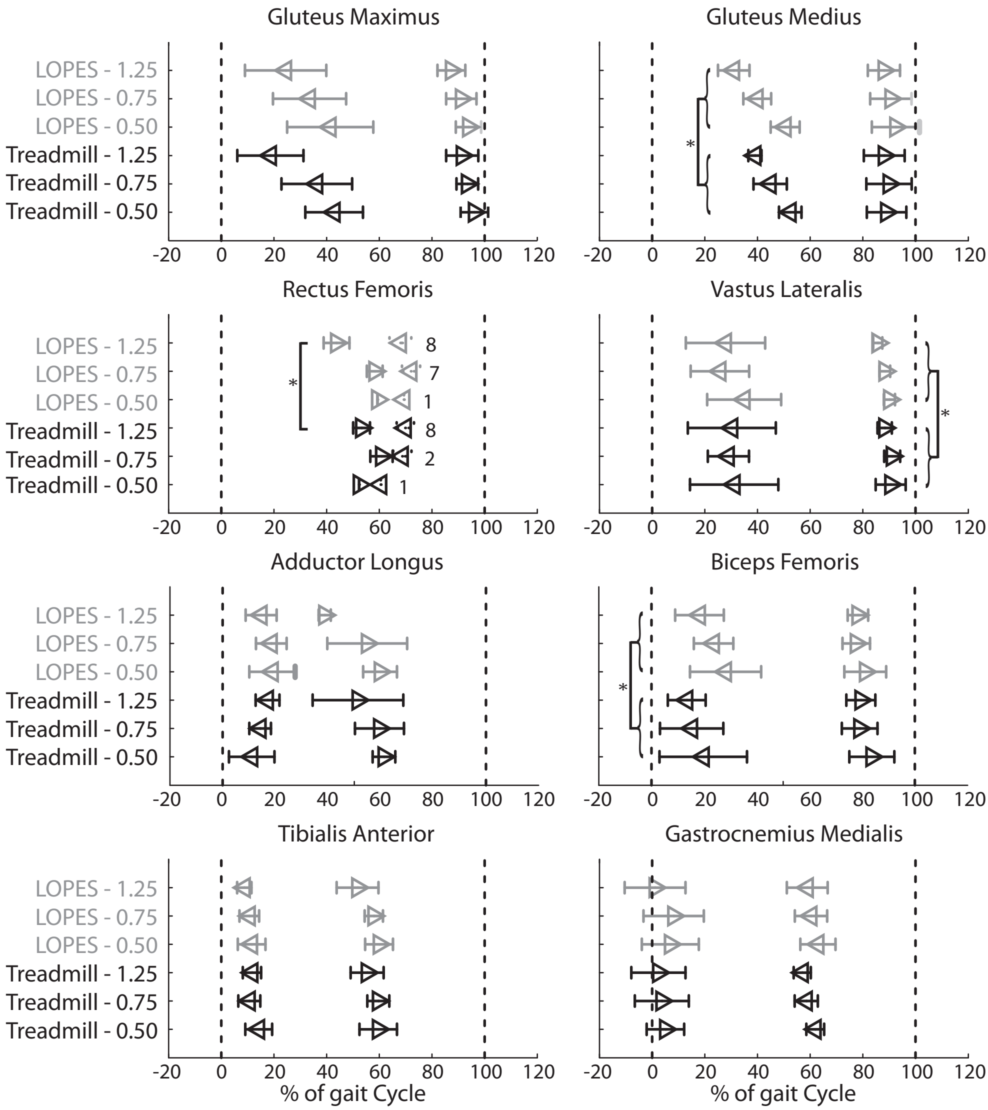

Figure 6.4. Mean and standard deviation of the onset and offset time of the main burst for each muscle. On the left side the conditions are indicated. The gray symbols indicate the different LOPES conditions and the black symbols the different treadmill conditions. The ▷ symbol indicates the onset of the burst and the ◁ symbol indicates the offset 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 a *.

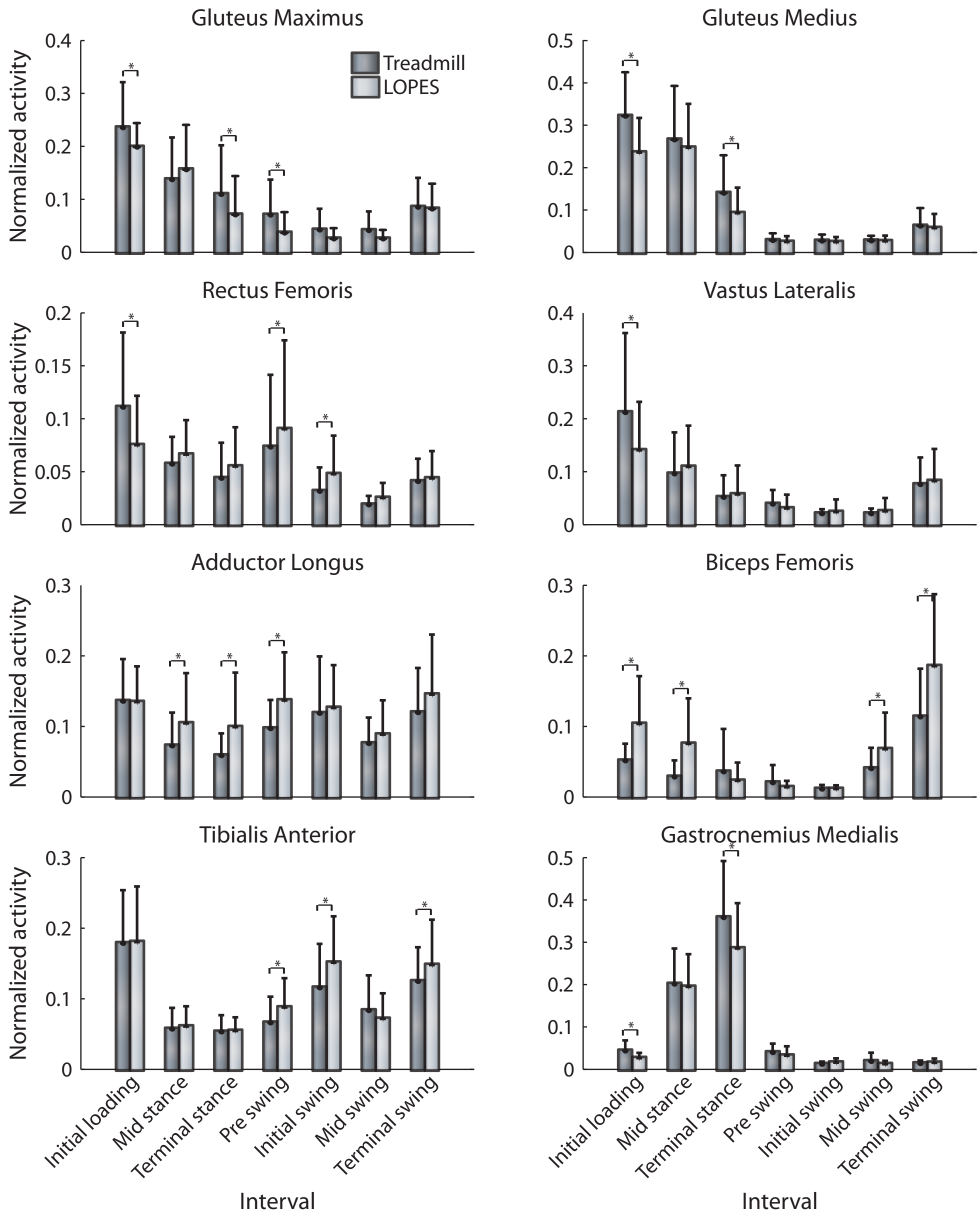

Figure 6.5. Mean normalized integrated activity for each muscle over the different phases for LOPES walking and treadmill walking.

The bi-articular biceps femoris showed an increase of its activity for LOPES. 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, while for the middle and lowest velocity the difference was not significant. The lower leg muscles both 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.

6.5 **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 Table 6.1 and 6.2) 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 (Hausdorff *et al.*, 1995; Leardini *et al.*, 2005; Cereatti *et al.*, 2006). 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 first 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.

6.5.1 **causes of observed difference**

pelvis motion

The amplitude of the muscle activity did show 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. Although the controller was set to regulate the sideward forces on the exoskeleton to zero, it 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 the connective constructional parts at the left and right leg. The aforementioned changes might have been aggravated by restraining the pelvic rotation, though the pelvic rotations during normal walking are generally rather small ((Perry, 1992)) and the fixation of the subject into device allowed some movement, which was not measured in this experiment. A similar study using the PAM (Aoyagi *et al.*, 2007), 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 have shown to play an important role in forward progression during mid stance and terminal stance, respectively (Neptune *et al.*, 2001; Gottschall & Kram, 2003; Neptune *et al.*, 2004). 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) had decreased, which could be the result of an overall decrease of the amplitude of the fluctuations in the acceleration and deceleration of the CoM through continuously opposing forces like friction (in the linear guide) and the inertia of the exoskeleton. During the experiments, we indeed observed relatively small movements of the pelvis while walking in Lopes, compared to free walking (Veneman *et al.*, 2007b). However the experimental setup did not allow measurement of the pelvis position in both conditions, so we were not able to directly compare them.

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 movement 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 6.1 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 Figure 6.3) 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 reported results in this study are in agreement with a study investigating the effect of an additional mass to the lower leg (Noble & Prentice, 2006). Adding a 2 kg weight to the left lower leg resulted in an increase of the hip extensor torque and an increase of the knee flexor torque at the end of the swing phase. Both of these torques can be generated by the bi-articular biceps femoris. In addition, they showed that initial swing was characterized by an increase of the hip flexion torque and knee extension torque. These results correspond to the increased rectus femoris activity during pre-swing and initial swing. Noble and Prentice(2006) also found 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 in the study of Noble and Prentice and in other studies (Skinner & Barrack, 1990) the addition of weight has shown to increase the swing time, which is also 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 the 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.

Apart of the changes in integrated activity of the muscles, some muscles also showed changes in timing of their activity. Changes in timing indicate that a different coordination is required for walking in the device. The observed changes were in some cases so small that it could be questioned whether the changes were also relevant. Based on results of Perry (Perry, 1992), Buurke (Buurke, 2005) 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 were 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.

comparison with other studies

Some other studies also compared the activation patterns while walking with a robotic device with the patterns outside the device (Aoyagi *et al.*, 2007; Schmidt *et al.*, 2007a). However, most of these studies restricted their analysis to a qualitative comparison and did not perform a detailed quantitative analysis as was done in this study. Only Hidler and Wall (Hidler & Wall, 2005) performed a similar detailed analysis when they assessed the alterations in muscle activation patterns when subjects walked 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 in this study.

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 (Veneman *et al.*, 2007b) 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 larger resemblance of EMG patterns in LOPES walking with free walking compared to Lokomat and free walking, should therefore be mainly contributed to the zero-impedance control that replaced the joint-trajectory (position) control.

In this respect it is important to note that the comparison of EMG patterns is quite cumbersome in a position controlled device. Despite that the position control is not ultimately stiff and that the patient fixation in the device will allow some movement, the generated muscle activity will only slightly influence the resulting motions, in other words, different activation pattern would result in approximately the same motion pattern. 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), ánd how natural these trajectories are. This combination makes outcomes hard to interpret. Of course, for healthy subjects, it will feel more natural to walk according to the device instead of working against it (Aoyagi *et al.*, 2007).

6.5.2 **implications for assist-as-needed**

Impedance controlled devices are especially suited to implement assist-as-needed control algorithms. In these algorithms the magnitude of the provided assistance can 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 case (Ekkelenkamp *et al.*, 2007; Van Asseldonk *et al.*, 2007), in which selective assistance is provided on several essential 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 those aspects of walking which are impaired. To decide which movements can be considered impaired and which not, reference values of healthy subjects are required. This study provided such reference values, which can be used to decide whether selective assistance is required.

One of the main deviations from normal walking was that maximal knee flexion during swing was decreased while walking with LOPES. Therefore, we should not expect that patients show a return to normal values when the assistance is progressively decreased to zero, instead we should be satisfied with a smaller knee flexion as this might be sufficient to show normal knee flexion outside the device.

Instead of adapting the reference values, the results could also be used to apply a controller based correction or an adaptation of the mechanical design such that the values achieved when no assistance is provided get closer to normal values. As the decreased knee flexion was likely the result of the increased inertia and weight of the swing leg, a further decrease of the weight of the exoskeleton or using a controller that corrects for this weight and inertia could bring the amount of knee flexion closer to normal values.

The results also indicated that there was some resistance in the horizontal pelvis motions. These degrees of freedom 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, when this assistance in balance control may be seriously needed. However, it is an undesirable effect when the subject needs to be encouraged to recover his balance control during gait through progressively decreasing the amount of provided assistance. Again, either mechanical or controlled based reduction of these disturbing effects may be useful. However, it is hard to predict how much gain we can achieve. The weight could possibly be substantially reduced, but especially the effect of adapting the controller is unpredicTable 6.as correcting for inertia would require the incorporation of acceleration or force sensors and this may result in unforeseen difficulties in the stability of the controller.

6.5.3 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, because task-specificity can be interpreted as: walking during training similar to natural walking. 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; the learned activation pattern would not be appropriate for walking over ground. As mentioned before, the patient-in-charge mode reflects the end stage of training as here (ideally) no assistive forces are applied by the robot that interfere with the patients' self-generated forces. Based on the results of this study in healthy subjects we can conclude that LOPES seems to have a good potential to provide a task specific training. It is however not known how close to normal walking training should be in order to consider it as sufficiently task-specific. This is an important issue that can be resolved only with realizing actual training sessions with patients.

As LOPES is going to be used to retrain stroke patients, we need to assess in future research 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 (Chen et al., 2005; Den Otter et al., 2006; Olney and Richards, 1996). This places additional demands on the degrees of freedom of the LOPES exoskeleton. For example, a large group of stroke patients shows an increased abduction during the swing phase (Kim and Eng, 2004). 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 possibly 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 optimal way of walking during training, which is not necessarily the healthy symmetrical way of walking (Buurke, 2005). When these movements can be made in LOPES, patients have some flexibility in developing their own walking pattern while training in LOPES, which might result in larger functional gains.

6.5.4 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 variability, which is an important advantage of impedance control. The changes in muscle activity could be attributed to the inertia of the device. As, in impedance controlled devices the device dynamics cannot be compensated by the controller, and will be perceptible for the user, improvements on this aspect can only be realised by decreasing the weight of the device.

The results indicate that LOPES, or in general an impedance controlled device with sufficiently low possible impedance and the right 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 a training that teaches skills transferable to free walking, remains a question that can only be answered by clinical research.

6.6 acknowledgements

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6.7 references

Agrawal SK, Banala SK, Fattah A, Sangwan V, Krishnamoorthy V, Scholz JR & Hsu WL. (2007). Assessment of motion of a swing leg and gait rehabilitation with a gravity balancing exoskeleton. *IEEE Transactions on Neural Systems and Rehabilitation Engineering* 15: 410-420

Aoyagi D, Ichinose WE, Harkema SJ, Reinkensmeyer DJ & Bobrow JE. (2007). A robot and control algorithm that can synchronously assist in naturalistic motion during body-weight-supported gait training following neurologic injury. *IEEE Transactions on Neural Systems and Rehabilitation Engineering* 15: 387-400

Bayona NA, Bitensky J, Salter K & Teasell R. (2005). The role of task-specific training in rehabilitation therapies. *Topics in Stroke Rehabilitation* 12: 58-65

Buurke JH. (2005). *Walking after stroke: Co-ordination patterns & functional recovery.* Enschede.

Cai LL, Fong AJ, Otoshi CK, Liang Y, Burdick JW, Roy RR & Edgerton VR. (2006). Implications of assist-as-needed robotic step training after a complete spinal cord injury on intrinsic strategies of motor learning. *The Journal of Neuroscience* 26: 10564-10568

Cereatti A, Della Croce U & Cappozzo A. (2006). Reconstruction of skeletal movement using skin markers: comparative assessment of bone pose estimators. *Journal of Neuroengineering and Rehabilitation* 3: 7

Colombo G, Joerg M, Schreier R & Dietz V. (2000). Treadmill training of paraplegic patients using a robotic orthosis. *Journal of Rehabilitation Research and Development* 37: 693-700.

Dobkin BH. (2004). Strategies for stroke rehabilitation. *Lancet Neurology* 3: 528-536

Ekkelenkamp R, Veltink PH, Stramigioli S & van der Kooij H. (2007). Evaluation of a VMC for the selective support of gait functions using an exoskeleton. In *Proceedings of ICORR 2007 - IEEE International Conference on Rehabilitation Robotics*: 693-699 Noordwijk, The Netherlands.

Ferraro M, Palazzolo JJ, Krol J, Krebs HI, Hogan N & Volpe BT. (2003). Robot-aided sensorimotor arm training improves outcome in patients with chronic stroke. *Neurology* 61: 1604-1607

Gottschall JS & Kram R. (2003). Energy cost and muscular activity required for propulsion during walking. *Journal of Applied Physiology* 94: 1766-1772

Hausdorff JM, Ladin Z & Wei JY. (1995). Footswitch system for measurement of the temporal parameters of gait. *Journal of Biomechanics* 28: 347-351

Hermens HJ, Freriks B, Merletti R, Stegeman D, Blok J, Rau G, Disselhorst-Klug C & Hägg G. (1999). *European Recommendations for Surface ElectroMyoGraphy* Roessingh Research and Development, Enschede, The Netherlands.

Hesse S, Uhlenbrock D & Sarkodie-Gyan T. (1999). Gait pattern of severely disabled hemiparetic subjects on a new controlled gait trainer as compared to assisted treadmill walking with partial body weight support. *Clinical Rehabilitation* 13: 401-410.

Hidler J, Nichols D, Pelliccio M & Brady K. (2005). Advances in the understanding and treatment of stroke impairment using robotic devices. *Topics in Stroke Rehabilitation* 12: 22-35

Hidler JM & Wall AE. (2005). Alterations in muscle activation patterns during robotic-assisted walking. *Clinical Biomechanics (Bristol, Avon)* 20: 184-193

Hogan N, Krebs HI, Rohrer B, Palazzolo JJ, Dipietro L, Fasoli SE, Stein J, Hughes R, Frontera WR, Lynch D & Volpe BT. (2006). Motions or muscles? Some behavioral factors underlying robotic assistance of motor recovery. *Journal of Rehabilitation Research and Development* 43: 605-618

Husemann B, Muller F, Krewer C, Heller S & Koenig E. (2007). Effects of locomotion training with assistance of a robot-driven gait orthosis in hemiparetic patients after stroke: a randomized controlled pilot study. *Stroke* 38: 349-354

Israel JF, Campbell DD, Kahn JH & Hornby TG. (2006). Metabolic costs and muscle activity patterns during robotic- and therapist-assisted treadmill walking in individuals with incomplete spinal cord injury. *Physical Therapy* 86: 1466-1478

Krebs HI & Hogan N. (2006). Therapeutic Robotics: A Technology Push. *Proceedings of the IEEE* 94: 1727-1738

Krebs HI, Palazzolo JJ, Dipietro L, Volpe BT & Hogan N. (2003). Rehabilitation robotics: Performance-based progressive robot-assisted therapy. *Autonomous Robots* 15: 7-20

Kwakkel G, van Peppen R, Wagenaar RC, Wood Dauphinee S, Richards C, Ashburn A, Miller K, Lincoln N, Partridge C, Wellwood I & Langhorne P. (2004). Effects of augmented exercise therapy time after stroke: a meta-analysis. *Stroke* 35: 2529-2539

Kwakkel G, Wagenaar RC, Koelman TW, Lankhorst GJ & Koetsier JC. (1997). Effects of intensity of rehabilitation after stroke. A research synthesis. *Stroke* 28: 1550-1556

Leardini A, Chiari L, Della Croce U & Cappozzo A. (2005). Human movement analysis using stereophotogrammetry. Part 3. Soft tissue artifact assessment and compensation. *Gait & posture* 21: 212-225

Lunenburger L, Colombo G & Riener R. (2007). Biofeedback for robotic gait rehabilitation. *Journal of Neuroengineering and Rehabilitation* 4: 1

Neckel N, Wisman W & Hidler J. (2006). Limb Alignment and Kinematics Inside a Lokomat Robotic Orthosis. In *Proceedings of EMBS 2006 28th Annual International Conference of the IEEE Engineering in Medicine and Biology Society*: 2698-2701. New York City, New York, USA

Neckel ND, Nichols D & Hidler JM. (2007). Joint moments exhibited by chronic stroke subjects while walking with a prescribed physiological gait pattern. In *Proceedings of ICORR 2007 - IEEE International Conference on Rehabilitation Robotics*: 771-775, Noordwijk, The Netherlands.

Nene A, Byrne C & Hermens H. (2004). Is rectus femoris really a part of quadriceps?: Assessment of rectus femoris function during gait in able-bodied adults. *Gait & Posture* 20: 1-13

Neptune RR, Kautz SA & Zajac FE. (2001). Contributions of the individual ankle plantar flexors to support, forward progression and swing initiation during walking. *Journal of Biomechanics* 34: 1387-1398

Neptune RR, Zajac FE & Kautz SA. (2004). Muscle force redistributes segmental power for body progression during walking. *Gait & Posture* 19: 194-205

Noble JW & Prentice SD. (2006). Adaptation to unilateral change in lower limb mechanical properties during human walking. *Experimental Brain Research* 169: 482-495

Perry J. (1992). *Gait Analysis*. Slack Inc, Ontario.

Pohl M, Werner C, Holzgraefe M, Kroczek G, Mehrholz J, Wingendorf I, Hoolig G, Koch R & Hesse S. (2007). Repetitive locomotor training and physiotherapy improve walking and basic activities of daily living after stroke: a single-blind, randomized multicentre trial (DEutsche GAngtrainerStudie, DEGAS). *Clinical Rehabilitation* 21: 17-27

Riener R, Lunenburger L, Jezernik S, Anderschitz M, Colombo G & Dietz V. (2005). Patient-cooperative strategies for robot-aided treadmill training: first experimental results. *IEEE Transactions on Neural Systems and Rehabilitation Engineering* 13: 380-394

Roetenberg D, Buurke JH, Veltink PH, Forner Cordero A & Hermens HJ. (2003). Surface electromyography analysis for variable gait. *Gait & Posture* 18: 109-117

Schmidt H, Volkmar M, Werner C, Helmich I, Piorko F, Krüger J & Hesse S. (2007a). Muscle activation patterns of healthy subjects during floor walking and stair climbing on an end-effector-based gait rehabilitation robot. In *Proceedings of ICORR 2007 - IEEE International Conference on Rehabilitation Robotics*: 1077-1084. Noordwijk, The Netherlands.

Schmidt H, Werner C, Bernhardt R, Hesse S & Kruger J. (2007b). Gait rehabilitation machines based on programmable footplates. *Journal of Neuroengineering and Rehabilitation* 4: 2

Skinner HB & Barrack RL. (1990). Ankle weighting effect on gait in able-bodied adults. *Archives of Physical Medicine and Rehabilitation* 71: 112-115

Staude GH. (2001). Precise onset detection of human motor responses using a whitening filter and the log-likelihood-ratio test. *IEEE Transactions on Biomedical Engineering* 48: 1292-1305

Teasell R, Bitensky J, Salter K & Bayona NA. (2005). The role of timing and intensity of rehabilitation therapies. *Topics in Stroke Rehabilitation* 12: 46-57

Van Asseldonk EHF, Ekkelenkamp R, Veneman JF, van der Helm FCT & van der Kooij H. (2007). Selective control of a subtask of walking in a robotic gait trainer(LOPES). In *Proceedings of ICORR 2007 - IEEE International Conference on Rehabilitation Robotics:* 841-848. Noordwijk, The Netherlands.

Van der kooij H, Veneman J & Ekkelenkamp R. (2006). Design of a compliantly actuated exo-skeleton for an impedance controlled gait trainer robot. In *Engineering in Medicine and Biology Society, 2006 Proceedings of the 28th Annual International Conference of the IEEE*: 189-193 New York City, USA.

Van Peppen RP, Kwakkel G, Wood-Dauphinee S, Hendriks HJ, Van der Wees PJ & Dekker J. (2004). The impact of physical therapy on functional outcomes after stroke: what's the evidence? *Clinical Rehabilitation* 18: 833-862

Veneman JF, Ekkelenkamp R, Kruidhof R, van der Helm FCT & van der Kooij H. (2006). A Series Elastic- and Bowden-Cable-Based Actuation System for Use as Torque Actuator in Exoskeleton-Type Robots. *The International Journal of Robotics Research* 25: 261-281

Veneman JF, Kruidhof R, Hekman EEG, Ekkelenkamp R, Van Asseldonk EHF & Van der Kooij H. (2007a). Design and Evaluation of the LOPES Exoskeleton Robot for Interactive Gait Rehabilitation. *IEEE Transactions on Neural Systems and Rehabilitation Engineering* 15: 379-386

Veneman JF, Van Asseldonk EHF, Ekkelenkamp R, van der Helm FCT & van der Kooij H. (2007b). Evaluation of the effect on walking of balance-related degrees of freedom in a robotic gait training device. In *Proceedings of ICORR 2007 - IEEE International Conference on Rehabilitation Robotics:* 868-875 Noordwijk, the Netherlands.

Von Zitzewitz J, Bernhardt M & Riener R. (2007). A novel method for automatic treadmill speed adaptation. *IEEE Transactions on Neural Systems and Rehabilitation Engineering* 15: 401-409

chapter 7

general discussion

7.1 introduction

This thesis had two aims, both related to the recovery mechanisms underlying restoration of motor function in stroke survivors. The first aim of this thesis was to develop and evaluate methods which can be used to distinguish between restitution and compensation in the recovery of function in the lower extremities. In our studies we focused on balance control and developed two different methods (chapter 2 and 3) which can be used to assess the function of each separate leg in stabilization of the body. One of these methods was successfully applied in a cross-sectional study and an effect study in stroke survivors (chapter 2 and 4). The potential of the other one was shown in the evaluation of balance responses in a healthy subject (chapter 3).

The second aim of this thesis was to provide a solid scientific basis for using assist-as-needed algorithms in robot-aided gait training with the new robotic gait trainer LOPES, which we developed. In assist-as-needed algorithms the provided robotic assistance is adapted to the capabilities of the subject. Ultimately this kind of algorithm can be used in gait training with LOPES that specifically emphasizes the use of original (restitution) or alternative (compensation) movement strategies. Based on the effectiveness of these interventions the importance of the different recovery mechanisms can be inferred. In this thesis, we provided a basis for using these algorithms. First, we showed the need to adapt the provided assistance to achieve optimal motor learning. Assisting forces reduced the rate and amount of learning (chapter 5). Second, we demonstrated that near-to-normal walking was possible in the LOPES gait training device. This is a requisite for implementing assist-as-needed algorithms to control the robot during gait training in stroke survivors.

In the following discussion, we will focus on these aims separately. Starting from the next paragraph, the different studies related to the first aim will be discussed in further detail. First, the major findings will be discussed, followed by the essential differences between the used methods, limitations of the different methods and future directions for further extending the methods. In the last paragraph of the first part, the implications of the presented results in understanding the recovery mechanism in balance control will be shortly described. In the second part of the discussion, we will elucidate why and how assist-as-needed algorithms should be used to train different movement strategies in the LOPES gait training device. From this perspective, the results of the two studies related to the second aim will be discussed.

7.2 methods to quantify restitution and compensation in balance control of stroke survivors

7.2.1 major findings

The main complicating factor in quantifying restitution and compensation in balance control is that both legs contribute to balance control whereas the individual function of each leg needs to be known to deduce whether recovery of function occurs. Therefore the activity generated in each leg which contributes to balance control needs to be quantified. In this thesis, two different methods were presented to quantify the contribution of the different joint torques in balance control. The first method (chapter 2) used closed loop system identification techniques to quantify the stabilizing mechanisms of each ankle, which were subsequently used to determine the relative contribution of each ankle to stabilizing the body. The evaluation of this method in a group of 8 chronic stroke survivors showed that the paretic leg contributed significantly less to balance control than the non paretic leg. Furthermore the contribution of the paretic leg to balance control was significantly smaller than the contribution to weight bearing. In the second method (chapter 3), we used induced acceleration analysis to quantify the contribution of leg, trunk and arm joint torques to body stabilization. This method also made it possible to determine the actual destabilizing and restabilizing effect that the platform movement had while disturbing balance. As a "proof of principle" the method was used to analyze the balance responses of a healthy subject. The results showed that the platform movement dominated the destabilization but also the restabilization of the body. The restabilizing effect of the platform deceleration even surpassed the restabilizing effect of the joint torques. Of the different joint torques, the ankle torque provided the largest contribution to stabilization. We have attempted to use this method to quantify the contribution of the torques in the ankle, knee and hip of the paretic and non paretic leg in stroke survivors to determine the function of each leg in stabilizing the body. However due to problems we encountered in measuring the ground reaction forces with sufficient accuracy this was impossible. These problems will be discussed in more detail later on. As a consequence, we only used the first method to evaluate the effects of a newly designed balance training (chapter 4) that emphasized the use of the paretic leg in training the ability to withstand perturbation. The training was applied in 4 chronic stroke survivors and all of them could withstand larger perturbations at the end of the training period. However, in none of the subjects this increase was accompanied by a clear restitution of function in the paretic ankle and only in one patient, the training led to improvements of the functional balance. These results indicated that either the restitution of function occurred in the other joints of the paretic leg or that patients had learned a compensatory strategy to better counteract the perturbations.

7.2.2 essential differences between the different methods

System identification (chapter 2) and induced acceleration analysis (chapter 3) can both be used to quantify the contribution of the joint torques to balance control. However, they are based on very different concepts. System identification quantifies the neural mechanisms underlying the balance responses, whereas the induced acceleration analysis can be regarded as a mechanical approach to clarify the effect of the exerted joint torques. System identification is a technique used to identify a dynamic system, like the balance control system. The human balance control system consists of sensors (such as muscle spindles, joint receptors, the visual and the vestibular system) from which an estimate of the body position is derived, a controller that calculates the required corrective torques to counteract the "sensed" deviation of erect stance and the muscles that generate the required torque. In our approach we identified the balance control system (total of sensors, controller and muscles) of the paretic leg and the non paretic leg. That is, we quantified the amplitude and the timing of the generated torque in each ankle in response to the "sensed" deviation of erect stance. Subsequently we determined how the paretic and non paretic balance control systems contributed to the total system. All in all, the contribution was obtained by calculating the relation between the deviation of erect stance and the generated torque. In induced acceleration analysis, we determined how the generated torques contributed to the accelerations of the Center of Mass of the human body. From these accelerations we deduce the effectiveness of the generated torques in restoring balance. By adding the induced accelerations of the different joints of each leg and comparing their respective magnitudes, this method could be used to determine the function of the paretic and non paretic leg in balance control. In short, when we compare the essence of both approaches, system identification can be used to identify the balance control system and may help us to better understand how the generation of joint torques by the balance control system is affected in post stroke hemiparesis. Induced acceleration analysis may be useful to determine the efficacy of the generated torques generated by the balance control system on stabilization of the human body. In this respect they can be regarded as complementary methods.

A major difference between the application of the different approaches in our studies was the difference in the number of joints that were incorporated in the analysis. Induced acceleration analysis was used to quantify the contribution of the torques in all the leg joints, whereas only ankle torques were used in the system identification approach. This difference in number of joints that were incorporated in the analysis can be reduced to the aforementioned difference in the nature of the approach. As induced acceleration analysis is a mechanical approach it can be easily extended to incorporate different joints as long as these joint torques can be accurately quantified with inverse dynamics. With system identification the underlying control mechanisms are quantified. Incorporating the action of other joints in this kind of analysis is a lot more complex, as it requires the definition of a different architecture of the balance controller. Extending the system identification method to incorporate the contributions of other joints is one of our future aims and will be discussed in more detail in the paragraph about future directions.

A second difference between both methods, is that induced acceleration analysis can not only be used to determine the contribution of generated joint torques, but also to assess the contribution of other externally applied or internally generated forces on destabilization and restabilization of the human body. This is not possible with system identification. Actually, determining the induced acceleration of the externally applied platform translation provided some very import insights (Chapter 3) about the validity and reliability of using platform translations to study balance responses. As expected, the platform acceleration dominated the destabilization of the human body. Yet, it also showed that the platform deceleration dominated the restabilization. This large restabilizing effect reduces the need for the subject to generate an efficient response to the initial disturbance. Furthermore, studying the complete human response to the acceleration is hindered because the deceleration and acceleration normally follow one another shortly, especially when subjects are repeatedly exposed to the same perturbations. For instance when the perturbations are part of the evaluation of a training regime (Marigold *et al.*, 2005; Vearrier *et al.*, 2005), subjects might learn to count on the restabilizing effect of the perturbation. This will not only limit the reliability but also the validity of using platform perturbations to assess the decreased balance control in patient populations, as the generated response could not only reflect the reactive balance control but also the ability of the subject to take advantage of the effect of the deceleration. In order to increase the validity and reliability of platform translations and to get a genuine and complete response to platform perturbation, the time between the acceleration and deceleration of the platform translation should be maximized, or the amplitude of the deceleration should be kept relatively small compared to the amplitude of the acceleration. We incorporatd the latter approach in the design of the platform training as described in chapter 4.

7.2.3 limitations of the approaches

Both methods can well be used in the analyses of balance control, however there are some limitations that need to be taken into account. One of the most important limiting factors in the general use of induced acceleration analysis or system identification, especially when this method is extended to include different joints, is the accuracy of the calculation of the joint torques with inverse dynamics. The major source of error in inverse dynamics during standing is the accuracy of the measured ground reaction forces. In studying balance responses, the generated torques are generally rather small compared to for example walking. As a consequence, the errors in the ground reaction force measurements due to non linearity, and crosstalk between the different forces and torques can have a relatively large influence on the calculated joint torques. We have attempted to use induced acceleration analysis to study the responses of stroke survivors to sudden platform perturbations. The hip joint torques that were generated to counteract the perturbation were generally smaller than 20 Nm. However, in calibration experiments we discovered that cross talk and non linearities in the force sensors that we used in this experiment could give horizontal force errors exceeding 10 N. Errors of this magnitude could possibly also have occurred during the experiment and would have accounted for errors in the hip joint torques of approximately 10 Nm, which was 50 % of the total generated torques. Obviously these errors were unacceptable. This experience taught us that for quantifying the generated joint torques in response to platform perturbations, dedicated force plates are needed that can measure the horizontal forces within an accuracy of 1 N regardless of the exerted vertical forces and or torques. This is especially true, when relatively small perturbations are used. This is generally done in studying postural control in stroke survivors where only small torques are needed to counteract the perturbations. When large perturbations are used like the ones used in chapter 3, larger hip joint torques are required and errors in the horizontal forces will have a relatively smaller influence. This together with the fact that an accurate force plate was used to measure ground reaction forces, resulted in an accurate estimation of the generated joint torques in the study of chapter 3.

When using induced acceleration analysis, it is of fundamental importance that the kinematic constraints are respected. In our model (chapter 3) we apply a kinematic constraint to the ankle because we assume that the ankle does not move with respect to the platform. This is a crucial assumption underlying the calculation of the induced accelerations without an incorporated foot model. Therefore this assumption always has to be verified. All the movements that occur between the ankle and the platform contribute to the induced accelerations of the error term. This error term also captures all other errors made in the measurements and calculations, such as the calculation error of the joint torques. As long as the induced accelerations of the error term are relatively small compared to the measured accelerations, the application of the kinematic constraint is justified and the joint torques are calculated accurately.

7.2.4 future directions and alternative applications

As mentioned before, we aim to extend the system identification approach to include the action of the hip and knee joint(s). In the current approach, we assumed that the stabilizing mechanisms of the human body solely generate a corrective ankle torque that is based on the deviation from upright stance. Chapter 3 and other studies (Alexandrov *et al.*, 2005; Creath *et al.*, 2005) showed that the ankle torque does provide a large contribution to balance control. Still, torques around other joints, especially those around the hip, are also believed to play a role (Runge *et al.*, 1999;

Park *et al.*, 2004). In fact, the role of these more proximal joints might even be larger in stroke survivors compared to healthy subjects, as the severity of motor impairment increases from the proximal to the distal joints (Adams *et al.*, 1990). Incorporation of additional joints changes the architecture of our model of the stabilizing mechanisms from a Single Input Single Output (SISO) to a Multiple Input Multiple Output (MIMO) system. A challenge in identifying this stabilizing MIMO system, is to define the appropriate inputs. The major question will be: what information do the stabilizing mechanisms use to generate the combination of joint torques to counteract deviation of the body position? The deviation of upright stance could still be used, however, also joint angles (Park *et al.*, 2004), segment angles, eigen movements (Alexandrov *et al.*, 2005) or a combination of these are plausible inputs. Determining the appropriate inputs is a first and crucial step in identifying the MIMO system of the stabilizing mechanisms. Another challenge is to define additional perturbations because a separate independent perturbation is required for each additional joint in the balance controller (Ljung, 1999). For example, in addition to the forward backward platform perturbation used in chapter 2, a disturbing force could be exerted at the pelvis. Future research will indicate whether different perturbations can be used to reliably estimate the stabilizing mechanisms at more proximal joints. Ultimately these stabilizing mechanisms can be used to determine the contribution of the additional joints to human body stabilization.

In this thesis we have shown that the paretic ankle of chronic stroke survivors showed a decreased role in balance control (chapter 2). We did not further investigate whether the disturbed active postural responses often reported in stroke survivors (Badke & Duncan, 1983; Di Fabio & Badke, 1988; Marigold *et al.*, 2004; Marigold & Eng, 2006) were responsible for this decreased function or whether it was due to changes in passive responses. By extending our system identification methods, we will gain more insight in the importance of the active and passive postural responses. By using system identification, it is possible to apportion the balance restoring torques to active and passive pathways. These pathways work in parallel to restore balance. In the active pathway, the central nervous system generates a corrective torque by altering the muscle activation, based on the afferent information from the muscle spindles and golgi tendon organs. In the passive pathway, the visco-elastic properties of the muscle, joint and surrounding tissues, instantly generate a torque to counteract any joint movement and/or muscle stretch. Both pathways can be affected to a different extent in stroke survivors. Changes in the mechanical properties of the fibers, for instance, affect the passive pathway (Friden & Lieber, 2003), whereas the often reported hyperactive stretch reflex in stroke survivors alters the active pathway. Recently, a method based on system identification was developed to investigate the integrity and functioning of these pathways in the dynamics of a relaxed ankle joint of hemiparetic spastic stroke subjects (Galiana *et al.*, 2005; Thajchayapong *et al.*, 2006). Whether similar procedures can be used when the ankle is used in a functional task (that is in balance control) has to be verified. If so, we can discern which of the pathways is mainly responsible for the decreased contribution of the paretic leg in balance control and we might be able to determine whether restitution and/or compensation can be attributed to adaptations of the active or passive pathway

The presented approaches to quantify the contribution of the separate joint torques to balance control not only provide insight into balance control of stroke survivors, but also have great potential in other patient populations that are characterized by a clear asymmetry in the impairments of both legs. We have recently applied the method to determine the dynamic balance contribution (chapter 2) to subjects with a prosthetic leg (Van der Kooij *et al.*, 2007b) and to subjects with Parkinson's disease (Van der Kooij *et al.*, 2007a). The subjects with a prosthetic leg clearly showed a smaller contribution of the prosthetic ankle compared to the contribution of the normal ankle. For this patient population it would be of special interest to extend the method and to also identify the stabilizing mechanisms of the (prosthetic) knee joint and hip joint. This could make it possible to detect the additional value of specific prosthetic components on the stabilizing mechanisms of the prosthesis. The pilot study on patients with Parkinson's disease showed that these patients also had clear asymmetries in their dynamic balance control. This result was quite remarkable as patients reported that they were not aware of this and had no subjective problems with stability and standing. Another important finding was that in both populations, like in stroke survivors, the asymmetries in dynamic balance contribution were larger than the asymmetries in weight bearing, which indicated that the contribution to balance control of a leg is not a reflection of the weight bearing on that leg.

In this thesis, we focused on determining the contribution of each leg to stabilization in the sagittal plane. The same methods might also be applicable to study the contribution of each leg in the frontal plane. This could increase our understanding of the nature and recovery of impaired frontal plane balance in stroke survivors (Kirker *et al.*, 2000; De Haart *et al.*, 2004). Before the methods can be applied to the frontal plane, several difficulties have to be overcome. For the system identification approach, this is the returning issue of adding actions of different joints to our model of the stabilizing mechanisms. As frontal plane balance is believed to be regulated from the hip joints (Rietdyk *et al.*, 1999), the need to include the actions of the hips in our model is even larger than in the sagittal plane. For induced acceleration analysis, the main issue is that we need to model the separate legs and that these legs together with the pelvis and the floor form a closed chain. The incorporation of this closed chain in the forward model can cause large errors in the accuracy of the calculated induced accelerations because the reduced mass matrix can get (close to) singular. Furthermore, for both methods hip (and other) torques need to be calculated accurately, which requires that the horizontal forces are measured accurately.

Induced acceleration might also be useful to better understand the contribution of the different joint torques to the execution of tasks other than balance control. Originally, induced acceleration analysis was used to quantify the contribution of the different joint torques to progression and weight bearing during walking (Neptune *et al.*, 2001; Zajac *et al.*, 2003). The method used in these studies was based on forward simulations. The accompanying heavy computational load limited a broad application of this method. The implementation of an analytical approach to calculate the induced acceleration during gait could greatly reduce this computational load (Hof & Otten, 2005). However, the application of an analytical approach in gait faces several difficulties. The main problem is that the definition of the required kinematic constraint is troublesome. In our application of induced acceleration analysis for balance control, we defined the kinematic constraint in the ankle, as we assumed that the ankle was not moving with respect to the underground. In gait, the ankle is stationary for a short time during midstance, but moves considerably during the remainder of the stance phase. As a consequence, we cannot apply a kinematic constraint to the ankle during gait, nor to any other point on the foot. We first have to find a solution for this problem, before we can use induced acceleration analysis to determine the efficacy of the joint torques in the paretic and non paretic leg. This would provide a detailed analysis of the function of the paretic and non paretic leg in walking, which subsequently can be used to assess the recovery mechanisms.

In conclusion, to get a more complete view of the disturbed postural balance control and of recovery mechanisms, our future aim is to extend the system identification method in three different ways: incorporate the action of the knee and hip joint, distinguish the active and passive pathways and analyze frontal plane balance. The use of induced acceleration analysis could be extended to assess the efficiency of joint torques in tasks such as gait and frontal plane balance control.

7.2.5 implications of the obtained results in understanding the recovery mechanism

The previous paragraphs have shown that there are several ways in which the methods presented in this thesis can be further improved and/or extended. Still, in their current form they can already be used to fill gaps that exist in the knowledge about the underlying recovery mechanisms in stroke survivors. Only by distinguishing between restitution of function in the affected or adaptive compensation in the non affected leg, the nature of the recovery can be assessed. In this respect we want to stress once more that restitution of function can only be derived with confidence when the contribution of the paretic leg is evaluated within the execution of the task. For instance, the paretic leg could show a change in the generated activity during balance control in the course of rehabilitation, however if this activity is not well coordinated it will not actually contribute to the execution of the task. Hitherto, the few studies that repeatedly assessed the generated activity in the paretic leg during recovery of balance control (Ikai *et al.*, 2003; De Haart *et al.*, 2004; Marigold *et al.*, 2005) have not directly related this activity to the execution of the task, as appropriate methods were not available. Both of the methods presented in chapter 2 and 3 were developed specifically for this purpose. When using these approaches, restitution would be expressed in a larger contribution of the paretic leg in balance control.

In this thesis a first step was made in using the method of chapter 2 in studying the recovery mechanisms. The method was used in the evaluation of a balance training in four chronic stroke survivors. During this training, patients counteracted perturbations consisting of surface translations in lateral direction and in anterior/posterior direction. The acceleration profile of the perturbations was such that these perturbation “pushed” the body to either their paretic side or to the back. The push to the paretic side elicited the paretic leg to contribute to counteracting the disturbance. Additionally, the therapist encouraged the patients to use their paretic leg as much as possible to restore balance. So, the training focused on increasing the ability to withstand perturbations by increasing the postural contribution of the paretic leg. During the course of the training, all patients showed a gradual increase in the magnitude of the perturbations they could withstand. None of the them demonstrated a consistent change in the paretic dynamic balance contribution, which means that we cannot infer the occurrence of either restitution or compensation in the ankles from this study. Possibly the training resulted in altered control at the paretic or non paretic knee and hip that resulted in a larger contribution of these joints to stabilization. However, the contributions of these joints could not be assessed within this study.

The previous study was performed in chronic stroke survivors. An interesting next step would be to use the methods in a longitudinal design to study the recovery mechanisms in the first months after stroke, when normally the largest amount of functional recovery occurs. In this respect, it is interesting to see whether we can provide direct support for either of the recovery mechanisms. If we can provide support for both recovery mechanisms it is important to know whether one mechanism results in better functional improvements than the other and whether the use of specific recovery mechanisms is related to the motor impairments of the patients. Although it seems logical that subjects with more severe impairments tend to make more use of compensatory movement strategies (Cirstea & Levin, 2000; Michaelsen *et al.*, 2001), this does not have to imply that these subject are not able to substitute these strategies for closer to normal strategies when they regain some control or force in the involved joints.

Although the importance of restitution and compensation can be best derived from repeated assessment during rehabilitation or a specific training intervention, cross sectional studies can also provide us with some important considerations. By relating the contribution of the paretic leg to functional scales, it can be deduced whether a good functioning paretic leg is required for a good task performance or whether compensatory strategies can be effectively used. We showed (chapter 2) that subjects with a small contribution of the paretic leg to balance control (< 20%) thus with a badly functioning paretic leg could still attain a good functional balance. These results were in accordance with those of Bowden and colleagues (2006) who showed that subjects with severe hemiparesis could still reach normal walking velocities while their paretic leg was only marginally contributing to propulsion. These results indicate the importance of functional compensatory strategies and put forward some intriguing questions. What determines whether subjects can make functional use of compensatory strategies to achieve a good task performance? Can the compensatory strategies that are most effective in promoting functional recovery be identified? Can subjects be trained in using these strategies to achieve a better functional performance?

7.3 role of assist-as-needed algorithms in robot-aided training of different movement strategies

The second aim of this thesis was to provide a basis for assist-as-needed control algorithms which allow the user to use different movement strategies during robot-aided training. Training with robotic devices has become increasingly popular as these devices can provide intensive and task specific training and reduce the physical workload for physical therapists. These two elements are believed to be most important in achieving optimal outcome in rehabilitation (Kwakkel et al., 1997; Van Peppen et al., 2004; Bayona et al., 2005; Teasell et al., 2005). The outcome could possibly be further improved by increasing the self-induced activity of the patients during training and by focusing training on specific movement strategies. The essential feature of assist-as-needed algorithms is that the amount of assistance is adapted based on the abilities of the subject. This means that the assistive forces are decreased when the subject is able to self generate a larger portion of the required force to complete the task. Furthermore, assist-as-needed algorithms can allow the flexibility in providing assistance such that the use of different movement strategies can be stressed. In the following paragraphs we will provide evidence in support of assist-as-needed algorithms obtained from this thesis and literature. Subsequently we will elucidate how these algorithms can be used in gait training for stroke survivors using the gait rehabilitation robot LOPES.

7.3.1 support for assist-as-needed training from motor adaptation studies

The beneficial effect of decreasing the amount of assistance could be deduced from chapter 5. There we showed that adaptation of reaching movements to a new visuomotor transformation occurred faster and more complete, when subjects received less assistive forces that reduced the error during practicing. Apparently, experienced errors during the execution of movements were used more effectively in adaptation than the felt assistive forces. Instead of using the assistive forces in adaptation, the subjects relied upon the forces to correctly perform their reaching movements. In fact, when the assistive forces totally restrained the movement to the optimal trajectory, such that the subjects experienced the correct movement over and over again during practicing, this did not result in a correct performance of this movement once the forces were turned off. Even the group that only received small assistive forces showed a decreased amount and rate of adaptation compared to the group that did not receive any assistive forces. All in all, it seems that the part of the movement that the robot takes over is only slowly taken over by the subject.

As a first step in increasing our understanding of the effect of providing assistance in motor relearning of stroke survivors, we studied the effect of assisting forces on motor adaptation in healthy subjects. Studying healthy subjects allowed us to unambiguously assess the effect of assistive forces on adaptation, as subjects were all capable to perform the task. If this study would have been performed in stroke survivors, the interpretation of the results would have been complicated by differences in motor performance of the stroke survivors as a consequence of the differences in motor deficits (Scheidt & Stoeckmann, 2007). The observed results are applicable to stroke survivors as long as the motor learning processes present in healthy subjects are partially preserved in stroke survivors. Recent studies have provided evidence in support of this premise (Takahashi & Reinkensmeyer, 2003; Scheidt & Stoeckmann, 2007). These studies showed that stroke survivors could adapt their reaching movement to compensate for an externally applied force perturbation. However, they were less effective in adapting reaches to the perturbations as they made less use of the prior perturbation sizes and experienced movement errors in the subsequent reach (Scheidt & Stoeckmann, 2007). Both studies showed that the ability to adapt improved with increased motor function of the paretic arm. In addition, the integrity of the arm proprioception was also related to the amount of adaptation (Scheidt & Stoeckmann, 2007). In short, these studies showed that stroke patients retain the capacity to adapt with their paretic limb. Although it is likely that the ability to adapt in one session is related to the ability to relearn movements during rehabilitation over a period of days or weeks , this relationship has not yet been investigated. Assessing this relationship would be the last step in justifying the use of results from adaptation studies in healthy subject to motor relearning in stroke survivors.

The results presented in chapter 5 could place us in a dilemma, because it showed that providing no assistance during practice results in optimal learning, while we are developing and implementing robots to provide assistance. Yet, stroke survivors need the provided assistance to make (accurate) movements possible. In the light of providing assistance to stroke survivors, the results can be interpreted as showing that the amount of assistance should be kept as low as possible to maximize a relearning effect. But how small can the assistance be set to still promote relearning? Should the provided assistance be just enough to let the patient successfully accomplish the task or should it be a little lower so that the patient just falls short? The latter option seems to be better with respect to the results of chapter 5, as an error in the execution of the movements appeared to be required to further improve in the subsequent attempts and to keep the patient from relying on the provided assistance. As soon as the patient improves his performance, the level of assistance can be further decreased, to keep the patient motivated. This approach is the one used in assist-as-needed algorithms.

7.3.2 evidence from literature for the efficacy of assist-as-needed in stroke rehabilitation

Assist-as-needed algorithms are increasingly used in the control of robotic rehabilitation devices (Krebs *et al.*, 2003; Hogan *et al.*, 2006; Emken *et al.*, 2007). These algorithms can only be implemented in robots with an interactive control, such as impedance control. Interactive control implies that the interaction between the robot and the subject is controlled. In impedance control, the amount of provided assistance is dependent on the deviation from a reference pattern and on the set value for the impedance, the lower the impedance the less assistive force. In other words, the performance of the subject and the settings of the controller determine the amount of assistance. When using impedance control as basis for an assist-as-needed algorithm, the set value for the impedance is made dependent on the subject's performance, which makes the provided assistance even more dependent on the performance of the subject.

Growing evidence shows that assist-as-needed algorithms are more efficient in promoting recovery. In the widely used arm rehabilitation device MIT-MANUS a control algorithm has been implemented that adjusts the amount of robot assistance based on different metrics of the reaching performance, such as the ability to initiate movement, the movement speed and the movement extent (Krebs *et al.*, 2003; Hogan *et al.*, 2006). By using this algorithm stroke survivors achieved a greater reduction of impairments and, equally important, in 30 % less practice movements. Another approach is used by Dewald and colleagues. In a series of experiments they have convincingly demonstrated that the abnormal coupling between shoulder and elbow movement limits the reaching performance of chronic stroke survivors (Beer *et al.*, 1999; Dewald & Beer, 2001; Beer *et al.*, 2004; Beer *et al.*, 2007; Sukal *et al.*, 2007). Particularly, the amount of shoulder abduction which serves to support the limb during reaching, limits the reaching range of motion (work area). Therefore, in their training protocol (Ellis *et al.*, 2007), the robotic device was used to support the limb during reaching and the amount of robotic arm support was progressively decreased when the support was no longer needed (that is when the reaching performance increased). Although only preliminary results are available, they indicate that the subjects who had to progressively increase their own limb support, showed a larger improvement in their reaching work area than the control subjects in which the robotic assistance was not adapted. The use of compensatory strategies was prevented in their assessment of the work area, so these results provided evidence for a larger amount of restitution when training with assist-as-needed.

The aforementioned studies applied assist-as-needed algorithms in training of the arms. So far these algorithms have not been evaluated in robotic rehabilitation of the lower extremities in human subjects. The only evidence in support of assist-as-needed algorithms in gait training was derived from an animal study (Cai *et al.*, 2006), which showed that spinal mice training with an assist-as-needed algorithm showed a larger recovery of stepping ability than mice training with a fixed trajectory.

7.3.3 future use of assist-as-needed in the control of LOPES

To test the efficacy of assist-as-needed algorithms in robotic gait training in stroke survivors, we are implementing these algorithms in the impedance controlled gait training device LOPES. Implementation of these algorithms 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; that is when the impedance is set to zero. This is especially important in the control of gait training devices, as walking consists of several subtasks that can be impaired to different degrees in stroke survivors. We only want to assist the affected functions or subtasks, meaning that the remaining gait functions should be left unhindered. In chapter 6 we evaluated whether unhindered walking was possible with the device. We showed that the basic gait parameters, the variability in these parameters and the pattern of joint movements and muscle activity while walking with LOPES resembled those of walking without LOPES. The amount of knee flexion and the amplitude of the bursts of several leg muscles did change. These changes could be mainly attributed to the increased inertia of the swing leg when wearing the LOPES exoskeleton and the experienced resistance in the horizontal pelvis movements due to the inertia of the whole exoskeleton. Although the LOPES exoskeleton was purposefully designed to be light weight for instance by keeping the heavy motors from the exoskeleton, the inertia was the main factor that prevented unhindered walking. Therefore, further weight reduction can be considered to be a main aspect for further improvement.

Especially the resistance in the pelvis movements could hinder an optimal implementation of assist-as-needed. The degrees of freedom at the pelvis were incorporated in the design of LOPES to be able to train balance control as an integral part of gait training and to require an active push off from the subject. Training of these aspects of walking could greatly enhance robotic gait training, as stroke survivors are well known to have a decreased lateral stability and an impaired ability to generate a push off force with the paretic limb (Olney *et al.*, 1991; Teixeira-Salmela *et al.*, 2001; Kim & Eng, 2004). Furthermore, improvement in balance control were shown to be more important than for instance improvements in leg strength to achieve improvement in walking ability (Kollen *et al.*, 2005). The use of an impedance controller in LOPES, makes it possible to train the stability during walking, as it can provide some assistance to the patient, while also requiring an active stabilization of the patient himself. The observed remaining resistance in the pelvis movement provides an undesirable and uncontrollable stabilizing effect on the body. This could especially be disadvantageous in the late stages of training when the support should be minimal to encourage the patient in achieving the optimal improvement in walking. However it could well be, that at this point in gait rehabilitation, the patient is already better of with training without the device.

Use of assist-as-needed in training of different movement strategies.

Assist-as-needed algorithms can not only be used to stimulate the patient to self generate the required activity, but they can also provide flexibility in the execution of movements to perform compensatory movement strategies. The importance of compensatory movement strategies in the functional recovery of gait was deduced from several studies in the project "Herstel van lopen na een CVA" (Research program on recovery of gait following stroke) (Huitema *et al.*, 2004; Buurke, 2005; Den Otter *et al.*, 2006). In addition, Kim & Eng (2004) have indicated that the use of compensatory movements can be a very effective strategy to achieve an optimal functional walking pattern. However, all the commercially available robotic gait trainers (mechanized gait trainer, Reha-Stim, Berlin, Germany; and Lokomat, Hocoma AG, Volketswil, Switzerland) focus their training on restitution of the original movement patterns. To allow or support compensatory strategies, the robotic device should have redundancy in the degrees of freedom and the control strategy should not be based on controlling of "normal" kinematic trajectories. The LOPES device provides this redundancy and the implemented controller could also allow compensatory strategies. For the control of LOPES a specific assist-as-needed algorithm is implemented that selectively controls subtasks of walking (Ekkelenkamp *et al.*, 2007; Van Asseldonk *et al.*, 2007) (the development and implementation of these controllers is part of the PhD project of Ralf Ekkelenkamp). In this kind of control, walking is divided in subtasks (like foot clearance, lateral balance control, weight support) that all have to be accomplished successfully to progress without falling. The performance on each of the subtasks can be evaluated and regulated separately. The patient can use different strategies to accomplish a certain subtask, for example the patient can use a hip circumduction strategy instead of regular knee flexion to get enough foot clearance. When the therapist believes that the prognostics for regaining knee flexion are not good, he can decide to provide the patient with support to learn a hip circumduction strategy. In this respect, it is important to gain more insight in the recovery characteristics of the different gait impairments early in stroke to provide the therapist with a basis for making these decisions. For instance, Huitema and colleagues (2004) demonstrated that stroke survivors who had a diminished knee flexion range at the start of rehabilitation, did not show recovery of this ability during rehabilitation. Based on these results, a therapist could decide that a patient with a diminished knee flexion should preferably be trained to use compensatory movements. All in all, training in LOPES with selective control of subtasks does not necessarily have to result in restitution of a normal symmetrical walking pattern, but can be aimed to achieve the most functional walking pattern for this particular patient.

The large flexibility in providing assistance during gait training in LOPES opens up different possibilities to investigate the most efficient way of robotic gait training. Using LOPES different research questions can be addressed: Does adaptation of the robotic assistance based on the performance of the subject (assist-as-needed) result in larger or faster improvements of walking capacity than keeping the amount of support constant during training? Do the gains in walking ability attained with robotic gait training exceed those of conventional gait training? Can compensatory strategies effectively be trained using LOPES? Does allowing or facilitating compensatory strategies during robot-aided gait training result in larger functional improvements? Are the extra degrees of freedom in the pelvis effective in training of balance control during walking?

LOPES cannot only provide the different training regimes that are required to answer these questions, but can also be used for evaluations. The integrated sensors in the exoskeleton can be used to record the movements and the interaction forces during training. Using these measurements the changes in the walking performance of the subject can be evaluated overtime. In addition, through the implementation of measurement protocols, LOPES can be used to perform objective measurements of motor functions in the lower extremities such as independent joint control and maximal muscle force. Ideally, measurements to derive compensation and restitution should also be made possible in the device, so that one device can perform the necessary training and evaluations to deduce the importance of the different recovery mechanisms in restoring motor function in stroke survivors.

7.4 references

Adams RW, Gandevia SC & Skuse NF. (1990). The distribution of muscle weakness in upper motoneuron lesions affecting the lower limb. *Brain* 113 (Pt 5): 1459-1476

Alexandrov A, Frolov A, Horak F, Carlson-Kuhta P & Park S. (2005). Feedback equilibrium control during human standing. *Biological Cybernetics* 93: 309-322

Badke MB & Duncan PW. (1983). Patterns of rapid motor responses during postural adjustments when standing in healthy subjects and hemiplegic patients. *Physical Therapy* 63: 13-20.

Bayona NA, Bitensky J, Salter K & Teasell R. (2005). The role of task-specific training in rehabilitation therapies. *Topics in Stroke Rehabilitation* 12: 58-65

Beer RF, Dewald JP, Dawson ML & Rymer WZ. (2004). Target-dependent differences between free and constrained arm movements in chronic hemiparesis. *Experimental Brain Research* 156: 458-470

Beer RF, Ellis MD, Holubar BG & Dewald JP. (2007). Impact of gravity loading on post-stroke reaching and its relationship to weakness. *Muscle & Nerve* 36: 242-250

Beer RF, Given JD & Dewald JP. (1999). Task-dependent weakness at the elbow in patients with hemiparesis. *Archives of Physical Medicine and Rehabilitation* 80: 766-772

Bowden MG, Balasubramanian CK, Neptune RR & Kautz SA. (2006). Anterior-posterior ground reaction forces as a measure of paretic leg contribution in hemiparetic walking. *Stroke* 37: 872-876

Buurke JH. (2005). *Walking after stroke: Co-ordination patterns & functional recovery.* Enschede.

Cai LL, Fong AJ, Otoshi CK, Liang Y, Burdick JW, Roy RR & Edgerton VR. (2006). Implications of assist-as-needed robotic step training after a complete spinal cord injury on intrinsic strategies of motor learning. *The Journal of Neuroscience* 26: 10564-10568

Cirstea MC & Levin MF. (2000). Compensatory strategies for reaching in stroke. *Brain* 123: 940-953.

Creath R, Kiemel T, Horak F, Peterka R & Jeka J. (2005). A unified view of quiet and perturbed stance: simultaneous co-existing excitable modes. *Neuroscience Letters* 377: 75-80

De Haart M, Geurts AC, Huidekoper SC, Fasotti L & van Limbeek J. (2004). Recovery of standing balance in postacute stroke patients: a rehabilitation cohort study. *Archives of Physical Medicine and Rehabilitation* 85: 886-895

Den Otter AR, Geurts AC, Mulder T & Duysens J. (2006). Gait recovery is not associated with changes in the temporal patterning of muscle activity during treadmill walking in patients with post-stroke hemiparesis. *Clinical Neurophysiology* 117: 4-15

Dewald JP & Beer RF. (2001). Abnormal joint torque patterns in the paretic upper limb of subjects with hemiparesis. *Muscle & Nerve* 24: 273-283.

Di Fabio RP & Badke MB. (1988). Influence of cerebrovascular accident on elongated and passively shortened muscle responses after forward sway. *Physical Therapy* 68: 1215-1220.

Ekkelenkamp R, Veltink PH, Stramigioli S & van der Kooij H. (2007). Evaluation of a VMC for the selective support of gait functions using an exoskeleton. In *Proceedings of ICORR 2007 - IEEE International Conference on Rehabilitation Robotics:* 693-699. Noordwijk, The Netherlands.

Ellis MD, Sukal T & Dewald JPA. (2007). Active abduction may be the key therapeutic factor for rehabilitation of reaching work area in stroke: preliminary results from an ongoing randomized controlled trial. *Program No 8012 2007 Neuroscience Meeting Planner San Diego, CA: Society for Neuroscience, 2007 CD-ROM*

Emken JL, Benitez R & Reinkensmeyer DJ. (2007). Human-robot cooperative movement training: Learning a novel sensory motor transformation during walking with robotic assistance-as-needed. *Journal of Neuroengineering and Rehabilitation* 4: 8

Friden J & Lieber RL. (2003). Spastic muscle cells are shorter and stiffer than normal cells. *Muscle & Nerve* 27: 157-164

Galiana L, Fung J & Kearney R. (2005). Identification of intrinsic and reflex ankle stiffness components in stroke patients. *Experimental Brain Research* 165: 422-434

Hof AL & Otten E. (2005). Assessment of two-dimensional induced accelerations from measured kinematic and kinetic data. *Gait & Posture* 22: 182-188

Hogan N, Krebs HI, Rohrer B, Palazzolo JJ, Dipietro L, Fasoli SE, Stein J, Hughes R, Frontera WR, Lynch D & Volpe BT. (2006). Motions or muscles? Some behavioral factors underlying robotic assistance of motor recovery. *Journal of Rehabilitation Research and Development* 43: 605-618

Huitema RB, Hof AL, Mulder T, Brouwer WH, Dekker R & Postema K. (2004). Functional recovery of gait and joint kinematics after right hemispheric stroke. *Archives of Physical Medicine and Rehabilitation* 85: 1982-1988

Ikai T, Kamikubo T, Takehara I, Nishi M & Miyano S. (2003). Dynamic postural control in patients with hemiparesis. *American Journal of Physical Medicine & Rehabilitation* 82: 463-469; quiz 470-462, 484

Kim CM & Eng JJ. (2004). Magnitude and pattern of 3D kinematic and kinetic gait profiles in persons with stroke: relationship to walking speed. *Gait & Posture* 20: 140-146

Kirker SG, Jenner JR, Simpson DS & Wing AM. (2000). Changing patterns of postural hip muscle activity during recovery from stroke. *Clinical Rehabilitation* 14: 618-626.

Kollen B, van de Port I, Lindeman E, Twisk J & Kwakkel G. (2005). Predicting improvement in gait after stroke: a longitudinal prospective study. *Stroke* 36: 2676-2680

Krebs HI, Palazzolo JJ, Dipietro L, Volpe BT & Hogan N. (2003). Rehabilitation robotics: Performance-based progressive robot-assisted therapy. 15: 7-20

Kwakkel G, Wagenaar RC, Koelman TW, Lankhorst GJ & Koetsier JC. (1997). Effects of intensity of rehabilitation after stroke. A research synthesis. *Stroke* 28: 1550-1556

Ljung L. (1999). *System identification: Theory for the user*. Prentice-Hall, Upper Saddle River.

Marigold DS & Eng JJ. (2006). Altered timing of postural reflexes contributes to falling in persons with chronic stroke. *Experimental Brain Research* 171: 459-468

Marigold DS, Eng JJ, Dawson AS, Inglis JT, Harris JE & Gylfadottir S. (2005). Exercise leads to faster postural reflexes, improved balance and mobility, and fewer falls in older persons with chronic stroke. *Journal of the American Geriatrics Society* 53: 416-423

Marigold DS, Eng JJ & Timothy Inglis J. (2004). Modulation of ankle muscle postural reflexes in stroke: influence of weight-bearing load. *Clinical Neurophysiology* 115: 2789-2797

Michaelsen SM, Luta A, Roby-Brami A & Levin MF. (2001). Effect of trunk restraint on the recovery of reaching movements in hemiparetic patients. *Stroke* 32: 1875-1883

Neptune RR, Kautz SA & Zajac FE. (2001). Contributions of the individual ankle plantar flexors to support, forward progression and swing initiation during walking. *Journal of Biomechanics* 34: 1387-1398

Olney SJ, Griffin MP, Monga TN & McBride ID. (1991). Work and power in gait of stroke patients. *Archives of Physical Medicine and Rehabilitation* 72: 309-314

Park S, Horak FB & Kuo AD. (2004). Postural feedback responses scale with biomechanical constraints in human standing. *Experimental Brain Research* 154: 417-427

Rietdyk S, Patla AE, Winter DA, Ishac MG & Little CE. (1999). NACOB presentation CSB New Investigator Award. Balance recovery from medio-lateral perturbations of the upper body during standing. North American Congress on Biomechanics. *Journal of Biomechanics* 32: 1149-1158

Runge CF, Shupert CL, Horak FB & Zajac FE. (1999). Ankle and hip postural strategies defined by joint torques. *Gait & Posture* 10: 161-170

Scheidt RA & Stoeckmann T. (2007). Reach adaptation and final position control amid environmental uncertainty after stroke. *Journal of Neurophysiology* 97: 2824-2836

Sukal TM, Ellis MD & Dewald JP. (2007). Shoulder abduction-induced reductions in reaching work area following hemiparetic stroke: neuroscientific implications. *Experimental Brain Research* 183: 215-223

Takahashi CD & Reinkensmeyer DJ. (2003). Hemiparetic stroke impairs anticipatory control of arm movement. *Experimental Brain Research* 149: 131-140

Teasell R, Bitensky J, Salter K & Bayona NA. (2005). The role of timing and intensity of rehabilitation therapies. *Topics in Stroke Rehabilitation* 12: 46-57

Teixeira-Salmela LF, Nadeau S, McBride I & Olney SJ. (2001). Effects of muscle strengthening and physical conditioning training on temporal, kinematic and kinetic variables during gait in chronic stroke survivors. *Journal of Rehabilitation Medicine* 33: 53-60.

Thajchayapong M, Alibiglou L, Lilaonitkul T & Mirbagheri MM. (2006). Mechanical abnormalities of the spastic ankle in chronic stroke subjects. In *Engineering in Medicine and Biology Society, 2006 Proceedings of the 28th Annual International Conference of the IEEE*: 3688-3691. New York City, USA.

Van Asseldonk EHF, Ekkelenkamp R, Veneman JF, van der Helm FCT & van der Kooij H. (2007). Selective control of a subtask of walking in a robotic gait trainer(LOPES). In *Proceedings of ICORR 2007 - IEEE International Conference on Rehabilitation Robotics*: 841-848 Noordwijk, The Netherlands.

Van der Kooij H, Van Asseldonk EH, Geelen J, Van Vugt JP & Bloem BR. (2007a). Detecting asymmetries in balance control with system identification: first experimental results from Parkinson patients. *Journal of Neural Transmission* 114: 1333-1337

Van der Kooij H, Van Asseldonk EHF & Nederhand M. (2007b). Detecting asymmetries in balance control with system identification: first experimental results from above knee amputees. In *Proceedings of ICORR 2007 - IEEE International Conference on Rehabilitation Robotics*: 1055-1062. Noordwijk, The Netherlands.

Van Peppen RP, Kwakkel G, Wood-Dauphinee S, Hendriks HJ, Van der Wees PJ & Dekker J. (2004). The impact of physical therapy on functional outcomes after stroke: what's the evidence? *Clinical Rehabilitation* 18: 833-862

Vearrier LA, Langan J, Shumway-Cook A & Woollacott M. (2005). An intensive massed practice approach to retraining balance post-stroke. *Gait & Posture* 22: 154-163

Zajac FE, Neptune RR & Kautz SA. (2003). Biomechanics and muscle coordination of human walking: part II: lessons from dynamical simulations and clinical implications. *Gait & Posture* 17: 1-17

summary

Since restoration of motor function in walking and stance is an important aspect in the rehabilitation of stroke patients, insight in the responsible recovery mechanisms is of great importance. Generally, two recovery mechanisms are distinguished: restitution and compensation. In restitution, the increase in motor function is attributed to a return of original movements and or function of the paretic leg, whereas in compensation the increase in motor function is ascribed to the emergence of new movement strategies or to secondary adaptations in the non paretic leg to compensate for the decreased motor abilities in the paretic leg. To assess the contribution of the different mechanisms in the recovery of motor function, we need to develop appropriate methods to quantify restitution and compensation in time. Besides, the contribution of the separate recovery mechanisms can also be derived from the effectiveness of therapeutic interventions that focus specifically on using original movement patterns (restitution) or alternative movement strategies (compensation). Robotic rehabilitation devices can well be used in emphasizing the use of certain movement strategies during training, as they can support the movements of the subject during practicing. However, practicing compensatory strategies in the device is only possible if the mechanical design of the device allows for it and if the control of the device provides the flexibility to the subjects in choosing their movement strategy. The latter requires the implementation of "assist-as-needed" algorithms that only provide assistance to the subject when it is needed to successfully fulfill the task.

The aim of the studies reported in this thesis was twofold

1. Develop and evaluate methods which can be used to distinguish between restitution and compensation in the recovery of function in the lower extremities of stroke survivors.
2. Provide a basis for the use of assist-as-needed algorithms which allow the flexibility to use different movement strategies during robot-aided training.

Appropriate methods for quantifying restitution and compensation in the recovery of balance control are not available. The major complicating factor is that both legs can contribute to balance control. Only by quantifying the separate contribution of each leg to balance control we can derive whether improvements in balance control need to be attributed to the paretic or non paretic leg. In this thesis two methods are presented and evaluated that can be used to assess the function of each leg in stabilizing the human body.

The ankle torques provide a large contribution to stabilization of the human body by correcting deviations from upright stance. The magnitude and timing of the corrective torque in each ankle in response to a deviation reflect the stabilizing mechanisms in balance control and indicate how the ankle contributes to balance control. In the first method, we identified these stabilizing mechanisms of the paretic and non paretic ankle by using closed loop system identification techniques. The stabilizing mechanisms were estimated in a cross sectional study in eight chronic stroke survivors from their measured ankle torques and sway in response to a continuous forward backward platform perturbation. Subsequently, to quantify the relative contribution of each leg in balance control, the contribution of the stabilizing mechanisms of each separate leg to the stabilization of the total body was determined. We showed that the contribution to balance control was significantly lower in the paretic than in the non paretic leg. Furthermore, the results demonstrated that the paretic contribution to balance control was significantly lower than the contribution to weight bearing of the paretic leg. This indicated that stroke survivors are more easily able to use their paretic leg for the fairly static task of weight bearing than for the highly dynamic task of balance control. Finally, a large relative contribution of the paretic leg in balance control turned out not to be a prerequisite for a good functional balance, which indicated that the non paretic leg can compensate for the loss of function in the paretic leg.

The second approach differed from the first approach, in that it also included joint torques other than the ankle and other internal or external forces and torques in the calculation of the contribution to stabilization of the body. The presented method is based on an analytic implementation of the induced acceleration analysis. In induced acceleration analysis the effect of every single force/torque on the acceleration of the Center of Mass of the total body can be determined. From the induced acceleration on the Center of Mass, the (de)stabilizing character of a force/torque can be derived. As a "proof of principle" we used this method to analyze the postural responses of a healthy subject in response to unpredictable forward and backward translations of a moving platform. The results showed that the contributions of the different joint torques and other forces/torques to stabilization could be determined reliably. Remarkably, the restabilizing effect of the platform deceleration exceeded the stabilizing effect of the joint torques. This stresses the need to be careful in interpreting a subject's response to a platform translation in which the platform deceleration follows directly on the acceleration. With induced acceleration analysis the contribution of the different forces and torques to stabilization can be accurately determined and as such it can also be applied to assess the contribution of the joint torques generated in the paretic and non paretic leg of stroke survivors.

From longitudinal application of both methods, we can derive whether the function of the paretic ankle in balance control changes and whether restitution or compensation occurs. This analysis was performed (using the first approach) as part of an evaluation of a newly designed balance training for stroke survivors. This training specifically aimed for increasing the ability to withstand perturbations by restitution of function in the paretic leg. In this training, subjects stood on a movable platform and platform perturbations were used to elicit a functional response in the paretic leg. To optimally challenge the subject the magnitude of the perturbations was progressively increased during the training. The four chronic stroke survivors who participated in the study all showed an increase in the perturbation magnitude they could withstand during training. Still, this increase could not be attributed to a restitution of function in the paretic leg in any of the subjects, as the paretic dynamic balance contribution did not show a consistent change. In one subject the training resulted in a clear increase of the functional balance and walking ability, whereas in the other subjects the training did not seem to have an additional effect. Although this training forced the subjects in using their paretic leg in balance responses, it was not effective in provoking restitution in the paretic ankle, at least not in this small subset of stroke patients.

The previous studies described the results that fulfilled the first aim of this thesis. The last two studies were conducted in light of the second aim.

The last decade several robotic devices have been developed to assist stroke survivors during movement rehabilitation. Still, it is largely unknown what kind of assistance results in the optimal relearning of movements. Even in healthy subjects it is not known how assisting forces influence motor learning and whether subjects can effectively use the provided assistance in learning to perform the required movement after different attempts. To better understand the patient's response to different kinds of assistance, we quantified the effect of different assisting forces when healthy subjects learned a visuomotor task. In this task subjects had to adapt their reaching movement to altered visual feedback of the hand position. Five groups of 10 subjects received assisting forces while practicing to reach in the visually distorted environment. The forces differed per group in magnitude and direction, such that the reaching errors during practice were either decreased or enlarged. The results showed that none of the groups that received assisting forces demonstrated faster or more complete adaptation compared to a group that did not receive any assistance. Actually, the more assistance the subjects received in performing their reaches and thus the smaller the execution errors made, the smaller the amount and rate of adaptation. These results indicated that healthy subjects use assisting forces less efficiently than their execution errors in adapting their movements. Instead of using the assisting forces to adapt their movements, they relied upon these forces in movement execution, which further slowed down the adaptation. Although, the results were obtained from healthy subjects, they were in concordance with the basic idea behind the implementation of assist-as-needed algorithms in robotic devices for motor rehabilitation. The basic idea behind these algorithms is that the amount of assistance should be lowered as soon as the performance of the stroke survivor improves, such that he/she is optimally stimulated to make the movements and reliance upon the assisting forces is prevented.

So far, training with robotic devices has aimed mainly at improving motor function through restoration of original movement strategies. It could well be that larger gains in motor function could be obtained when also compensatory movements are allowed or even trained with the robotic device. Training alternative movement strategies puts additional demands on the mechanical design and the controller of the device. The mechanical designs should incorporate redundancy in the degrees of freedom and the controller should allow flexibility in how the movements are performed.

The robotic gait trainer called LOPES, that we have developed at our department, provides this required redundancy and flexibility. The flexibility is achieved by implementation of an assist-as-needed algorithm. Implementation of these algorithms 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 10 healthy subjects, while walking with LOPES attached to their pelvis and limbs and while walking freely on a treadmill. Overall, the patterns of the joint and segment movements and those of muscle activity while walking with LOPES, resembled those of free walking. However, a detailed quantitative analyses revealed several changes. The main changes were a decreased knee flexion during the swing phase, an increase of the muscle burst activity responsible for accelerating and decelerating the swing leg and a small decrease of the muscle burst activity involved in the push off. These differences could be mainly ascribed to the inertia of the device. All in all, the walking pattern with the device was similar to the normal walking pattern, but subjects adapted their muscle activity in order to achieve this. The next step will be to investigate whether stroke survivors can also walk fairly unhindered in the device and whether they can use their normal movement strategies. Subsequently, we aim to specifically train compensation strategies or original movements (restitution) in stroke survivors by using a specific implementation of an assist-as-needed algorithm that solely assists those subtasks of walking that are affected

samenvatting

Een beroerte, ook wel CerebroVasculair Accident (CVA) genoemd, leidt vaak tot spierzwakte en verminderde controle van bewegingen aan één zijde van het lichaam. Herstel van motorische taken zoals lopen en balans controle is een belangrijk aspect binnen de revalidatie van CVA patiënten. Dat betekent dat inzicht in de mechanismen die aan dit herstel ten grondslag liggen van groot belang is. Er kunnen twee herstelmechanismen worden onderscheiden: restitutie en compensatie. Restitutie houdt in dat een verbetering in de loopvaardigheid of balans controle het resultaat is van het herstel van de oorspronkelijke bewegingspatronen en/of functie in het aangedane been. Van compensatie is sprake wanneer de verbetering is toe te schrijven aan het ontstaan van nieuwe bewegingspatronen en/of aan aanpassingen in het niet-aangedane been om te compenseren voor de afgenomen functie in het aangedane been. Om het inzicht in het aandeel dat de verschillende mechanismen hebben in het herstel te vergroten zijn experimentele methodes nodig waarmee compensatie en restitutie in de tijd kunnen worden gekwantificeerd.

Het belang van de herstelmechanismen kan ook worden afgeleid uit de effectiviteit van trainingen die zich specifiek richten op het gebruik van de oorspronkelijke (restitutie) of alternatieve (compensatie) bewegingspatronen. Robotische revalidatieapparaten zijn zeer geschikt om het gebruik van bepaalde bewegingspatronen specifiek te trainen, aangezien zij de bewegingen van patiënten kunnen ondersteunen. Het oefenen van compensatie bewegingen is echter alleen mogelijk wanneer het mechanische ontwerp dit toelaat en wanneer de ondersteuning van het apparaat de flexibiliteit biedt aan de persoon om zijn eigen bewegingspatroon te kiezen. Voor dit laatste dient de robot te worden aangestuurd met een "ondersteuning-naar-behoefte" algoritme. In dit soort algoritmes biedt de robot alleen ondersteuning wanneer de patiënt deze nodig heeft om de motorische taak te voltooien.

Dit proefschrift heeft de volgende twee doelstellingen
1. Ontwikkelen en evalueren van methodes die gebruikt kunnen worden om restitutie en compensatie in het functionele herstel in de benen van CVA patiënten te onderscheiden.
2. Bieden van een onderbouwing voor het gebruik van ondersteuning-naar-behoefte algoritmes, die flexibiliteit toelaten in het gebruik van verschillende bewegingspatronen tijdens robotisch ondersteunde therapie.

Voor het kwantificeren van restitutie en compensatie in het herstel van de balans controle zijn geen geschikte methodes beschikbaar. De belangrijkste complicerende factor is dat beide benen kunnen bijdrage aan de balans controle of met andere woorden aan de stabilisatie van het lichaam. Alleen wanneer de bijdrage van elk afzonderlijk been kan worden gekwantificeerd, kan worden bepaald of verbeteringen in de balans controle dienen te worden toegeschreven aan het aangedane of niet-aangedane been. In dit proefschrift worden twee methodes geïntroduceerd en geëvalueerd die kunnen worden gebruikt om de functie van elk been in de balans controle te bepalen.

De krachten die rond de enkel worden gegenereerd leveren een belangrijke bijdrage aan de stabilisatie van het lichaam door een uitwijking van het lichaam uit de evenwichtsstand te corrigeren. De grootte en timing van deze corrigerende krachten in elk van de enkels in response op een bepaalde uitwijking geeft aan hoe ieder van de enkels bijdraagt aan de balans controle. Deze relatie tussen corrigerende krachten en uitwijking uit de evenwichtsstand noemen we de stabilisatie mechanismen. In de eerste methode hebben we de stabilisatie mechanismen van de aangedane en niet-aangedane enkel bepaald door gebruik te maken van gesloten lus systeem identificatie technieken. Dit werd gedaan in een cross sectioneel onderzoek waaraan 8 chronische CVA patiënten deelnamen. De stabilisatie mechanismen werden geschat met behulp van de gemeten krachten rondom de enkel en de lichaamsbeweging in reactie op een voortdurende platformverstoring in voor/achterwaartse richting. Vervolgens werd de bijdrage van de afzonderlijke benen in de balans controle gekwantificeerd door de bijdrage van de stabilisatie mechanismen van elk afzonderlijk been aan de stabilisatie van het totale lichaam te bepalen. De resultaten toonden aan dat de bijdrage van de aangedane enkel aan de balans controle significant kleiner is dan de bijdrage van het niet-aangedane been. Bovendien was de bijdrage aan de balans controle van het aangedane been significant lager dan de bijdrage van dit been aan de gewichtsondersteuning. Dit resultaat laat zien dat CVA patiënten hun aangedane been beter kunnen gebruiken voor de statische taak van gewichtsondersteuning dan voor de dynamische taak van balans controle. Tenslotte bleek een relatief grote bijdrage van het aangedane been in de balans controle geen vereiste voor een goede functionele balans, wat aantoont dat het niet-aangedane been kan compenseren voor de verloren functie in het aangedane been.

In de tweede methode werd niet alleen de bijdrage van de krachten rondom de enkel aan de stabilisatie van het lichaam bepaald maar ook die van de krachten rondom de andere gewrichten en van al de andere interne en externe krachten. In deze methode werden de "geïnduceerde versnellingen" van het lichaamzwaartepunt op een analytische wijze berekend. Uit deze geïnduceerde versnellingen kan het (de)stabiliserende effect van een bepaalde kracht worden afgeleid. De bruikbaarheid van deze methode hebben we geëvalueerd door deze methode te gebruiken in de analyse van balans correcties van een gezonde persoon in reactie op onverwachte voorwaartse en achterwaartse verstoringen met een beweegbaar platform. De resultaten toonden aan dat de bijdragen van de krachten rondom de gewrichten en van de andere krachten op de stabilisatie van het lichaam betrouwbaar konden worden bepaald. Een opmerkelijk resultaat was dat het stabiliserende effect van de platformvertraging groter was dan het stabiliserende effect van de krachten die de proefpersoon zelf had gegenereerd om de verstoring tegen te gaan. Dit geeft aan dat men voorzichtig moet zijn bij het interpreteren van de reactie van een persoon op een platformverstoring waarin de platformvertraging direct volgt op de versnelling. Aangezien geïnduceerde versnellingen analyse kan worden gebruikt om nauwkeurig de bijdrage van de verschillende krachten aan de stabilisatie te bepalen, kan deze analyse ook worden gebruikt om de bijdrage van de gegenereerde krachten in het aangedane en niet-aangedane been van CVA patiënten te bepalen.

De hiervoor beschreven methodes kunnen beide worden gebruikt in longitudinale of effect studies om te bepalen of de functie van het aangedane been in de balans controle in de tijd verandert en dus of er restitutie of compensatie optreedt. De eerste methode hebben we gebruikt als onderdeel van een evaluatie van een nieuw ontwikkelde balanstraining voor CVA patiënten. De gebruikte training richtte zich op het vergroten van het vermogen om verstoringen tegen te gaan door de functie van het aangedane been te verbeteren. In deze training stonden de patiënten op een beweegbaar platform en werden platform bewegingen gebruikt om een functionele reactie in het aangedane been uit te lokken. De grootte van deze verstoringen nam gedurende de trainingsperiode toe zodat de persoon optimaal werd uitgedaagd tijdens de training. De vier chronische CVA patiënten die deelnamen aan de studie lieten allen een toename zien van de grootte van de verstoring die ze konden weerstaan tijdens de training. Deze toename kon echter voor geen enkele patiënt worden toegeschreven aan een restitutie van functie in het aangedane been, aangezien de bijdrage van het aangedane been aan de balans controle geen consistente veranderingen liet zien. Bij één patiënt liet de training een duidelijke toename zien van de functionele balans en de loopvaardigheid, terwijl bij de overige patiënten de training geen toegevoegde waarde leek te hebben. Hoewel deze training de patiënten dwong om hun aangedane been te gebruiken in de balans reacties, hebben we niet kunnen aantonen dat dit een effectieve methode was om restitutie in de aangedane enkel te veroorzaken, tenminste niet in deze kleine groep van CVA patiënten.

Het voorgaande beschreef de resultaten die betrekking hadden op de eerste doelstelling van dit proefschrift. De volgende twee studies zijn gerelateerd aan de tweede doelstelling.

In het laatste decennium zijn verschillende revalidatierobots ontwikkeld die de bewegingen van CVA patiënten kunnen ondersteunen om zo het herstel te bevorderen. Toch is het nog grotendeels onbekend welke ondersteuning resulteert in het optimaal herleren van de bewegingen. Zelfs voor gezonde personen is het onbekend hoe ondersteunende krachten het motorisch leren kunnen beïnvloeden en of de personen de aangeboden ondersteuning tijdens het oefenen effectief gebruiken in het aanleren van de bewegingen. Om de reactie van de patiënt op verschillende vormen van ondersteuning beter te kunnen begrijpen, hebben we bij gezonde personen het effect van verschillende ondersteunende krachten onderzocht op het leren van een nieuwe motorische taak. Vijf groepen bestaande uit tien personen ontvingen ondersteunende krachten terwijl zij de nieuwe taak oefenden. De krachten verschilden per groep in grootte en richting, zodat de fouten in de uitvoering van de taak tijdens het oefenen werden vergroot of verkleind. De resultaten toonden aan dat geen van de groepen die ondersteunende krachten ontvingen de taak sneller of beter leerden dan een groep die geen ondersteuning ontving. Hoe meer ondersteuning de proefpersonen ontvingen en dus hoe minder fouten er werden gemaakt in de bewegingsuitvoering des te kleiner de mate en de snelheid van leren. Deze resultaten tonen aan dat gezonde personen voor het aanleren van hun bewegingen minder efficiënt gebruik maken van ondersteunende krachten dan van fouten in de uitvoering van de beweging. In plaats van de ondersteunende krachten te gebruiken in het leren vertrouwden de proefpersonen op de krachten in het goed uitvoeren van de bewegingen, wat het leren verder afremde. Hoewel deze resultaten zijn verkregen bij gezonde personen zijn ze in overeenstemming met het basisprincipe achter de implementatie van ondersteuning-naar-behoefte algoritmes in revalidatierobots. Het basisprincipe achter deze algoritmes is dat de mate van ondersteuning moet worden verlaagd zodra de bewegingsuitvoering van de CVA patiënt verbetert, zodat hij/zij altijd optimaal word gestimuleerd om de beweging zelf uit te voeren en zodat hij/zij niet leert te vertrouwen op de ondersteunende krachten.

Tot nu toe had het gebruik van revalidatierobots voornamelijk tot doel de motorische functie te verbeteren door restitutie van de oorspronkelijke bewegingen. Het is goed mogelijk dat grotere functionele verbeteringen kunnen worden bereikt wanneer ook compensatie strategieën worden toegestaan of zelfs getraind met de robots. Het trainen van alternatieve bewegingspatronen stelt extra eisen aan het mechanische ontwerp en de aansturing van het apparaat. Het mechanische ontwerp moet een redundant aantal vrijheidsgraden bezitten, wat inhoudt dat verschillende bewegingspatronen voor het uitvoeren van een taak mogelijk zijn. Bovendien moet de aansturing ook flexibiliteit toestaan in de manier waarop bewegingen worden uitgevoerd.

De revalidatie looprobot die wij binnen onze vakgroep Biomedische Werkthuigbouwkunde hebben ontwikkeld, LOPES genaamd, heeft deze redundantie en de vereiste flexibiliteit. De flexibiliteit wordt toegestaan door een specifieke implementatie van een ondersteuning-naar-behoefte algoritme. Implementatie van deze algoritmes vraagt niet alleen dat de robot de ondersteuning moet kunnen bieden wanneer dat nodig is, maar ook dat de robot de bewegingen niet moet hinderen wanneer de persoon geen ondersteuning nodig heeft. Als eerste stap in de implementatie van LOPES in looptrainingen, hebben we deze eigenschap van het apparaat geëvalueerd. Dit hebben we gedaan door de loopparameters, bewegingen en spieractiviteit van 10 gezonde proefpersonen te vergelijken wanneer zij liepen met LOPES bevestigd aan hun bekken en benen met wanneer zij vrij liepen op een lopende band. Over het algemeen leken de patronen van de gewrichtsbewegingen en van de spieractiviteit tijdens lopen met LOPES op die van vrij lopen. Een gedetailleerde kwantitatieve analyse liet echter verscheidene verschillen zien. De belangrijkste verschillen waren een afname van de knie flexie tijdens de zwaaifase, een toename van de spieractiviteit die verantwoordelijk is voor het versnellen en vertragen van het zwaaibeen, en een kleine afname van de spieractiviteit die verantwoordelijk is voor de afzet. Deze verschillen konden voornamelijk worden toegeschreven aan de extra massatraagheid van het apparaat. Kortom, het looppatroon in het apparaat was vergelijkbaar met het normale looppatroon, maar proefpersonen pasten de mate van activatie van hun spieren aan om dit te bereiken. De volgende stap is om te onderzoeken of CVA patiënten ook ongehinderd in het apparaat kunnen lopen en of zij hun normale loopstrategieën kunnen gebruiken. Vervolgens, hebben we tot doel om specifiek compensatie strategieën en originele strategieën (restitutie) te trainen bij CVA patiënten door een implementatie van een ondersteuning-naar-behoefte algoritme. Dit algoritme richt zich op op het enkel ondersteunen van die subtaken van het lopen die zijn aangedaan.

dankwoord

En dan ben ik nu toch aan het einde van mijn promotie gekomen. Vaak grijpen Brabanders een einde aan om welgemeend en met veel respect "Houdoe en bedankt" te zeggen, al dan niet gevolgd door "olé, olé". En dat wil ik bij deze dan ook doen! Maar goed, zo makkelijk wil ik me er niet van afmaken en een aantal mensen wil ik hier dan ook in een ware polonaise van bedankjes betrekken.

Aan de kop van deze polonaise lopen vol energie de altijd goed gemutste en inspirerende motors van het LOPES project, mijn promotor Frans van der Helm en co-promotor Herman van der Kooij. Beste Frans, tijdens je tweewekelijkse bezoeken aan Enschede moest er in korte tijd altijd veel overlegd worden. Toch slaagde je er altijd in de essentie van het geheel eruit te halen en goede adviezen te geven. Ik heb erg veel geleerd van de wijze waarop je problemen aanpakt en je wijze van redeneren. Bovendien was het van begin tot eind leuk om met je te werken en hebben we veel discussies gevoerd over voetbal. Alhoewel discussies een groot woord is, het was vaak meer het wederzijds inwrijven van de successen van de eigen club en de blamages van de ander. Hoewel ik me besef dat het staafdiagram in figuur 1 tot "represailles" gaat leiden, verwijs ik je toch graag naar de weegegeven feiten in deze figuur.

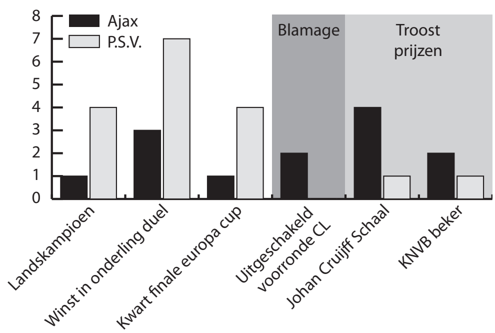

Figure D1. Successen en blamages van PSV en Ajax tijdens mijn promotie

Beste Herman, ik vond het erg leuk dat jij de afgelopen jaren mijn dagelijkse begeleider was. We delen een interesse voor dezelfde onderzoeksgebieden en jouw enorme enthousiasme wakkerde vaak mijn interesses nog verder aan. Ik heb erg veel waardering voor je kennis van zaken, je vastberadenheid en de manier waarop je altijd weer met een inventieve oplossing komt, al heb ik wel geleerd dat je jouw ideeën met een hoogdoorlaat filter moet bewerken (>2x per week). En niet te vergeten wilde je altijd graag meehelpen met de toch wel meesterlijke grappen die we verzonnen. Nu op naar een succesvolle implementatie van LOPES!

Voordat ik verder ga met de rest van mijn collega's wil ik al de mensen bedanken die deel hebben genomen aan de verschillende experimenten. Zonder de vrijwillige deelname van vele "proefpersonen" waren de studies in dit proefschrift niet mogelijk geweest. Ik besef dat de experimenten lang niet altijd even leuk waren, maar ik ben jullie dankbaar voor jullie geduld en inzet. In het bijzonder wil ik de CVA patiënten bedanken. Ik denk nog vaak met plezier terug aan de trainingen en metingen die voor jullie ongetwijfeld een zware belasting vormden maar toch altijd in een leuke sfeer plaatsvonden.

Samen met 2 andere promovendi, Jan Veneman en Ralf Ekkelenkamp, werkte ik aan het LOPES project, een waar multidisciplinair team waarin ieder van ons zijn eigen expertise had. Vele jaren zaten we samen op een kamer die later werd omgedoopt tot "De Gouden Kooij" (na onze illustere leider). Aan het begin vlogen de technische termen me vaak om de oren, maar inmiddels heb ik het gevoel dat ik een aardig woordje "technisch" spreek. Jan, elke keer sta ik er weer van te kijken van hoe nuchter jij alle dingen benaderd en beargumenteerd. Maar ja dat krijg je als je een ingenieur, filosoof en Fries combineert in één persoon. Wat je dan ook krijgt is een voetballer die op onorthodoxe wijze grote kansen afmaakt maar op even onorthodoxe wijze mist. Eigenlijk zou ik minstens een hele alinea nodig hebben om al mijn dank aan je op te kunnen schrijven maar je vakantie is toch al geboekt dus dat laat ik maar achterwege. Ralf jij leerde me het belang van plusjes en minnetjes en stond altijd klaar om te helpen. Op het eind van het project hebben we intensief samengewerkt in de uitvoering en de analyse van metingen en dat ging niet zonder slag of stoot, maar we hebben een mooi resultaat neergezet. Ook voor jou komt het einde in zicht en ik hoop dat ook jij het project met een feestje af zult sluiten. Zoals voor vele andere processen in het leven bleek ook in dit project een vrouwelijke inbreng op een gegeven moment noodzakelijk. Ons vaste team werd versterkt door Heike Vallery. Heike in je korte bezoekjes aan Enschede slaagde je er altijd in enorme hoeveelheden werk te verzetten voor de aansturing van LOPES, al moest je daar wel een dieet van chocomelk voor volgen. Hoewel ik het echt ontzettend knap vind hoeveel talen je spreekt, moet ik toch nog even iets kwijt. Het is "openbaar" dat ik je een heel gezellige collega vind, die ook nog eens erg betrokken is, maar het analyseren van "rouwe" data gaat me toch iets te ver.

En dan zijn er natuurlijk ook nog de andere collega's van de vakgroep Biomedische Werktuigbouwkunde, jullie hebben me voorzien van goede tips, hebben gezorgd dat ik mijn frustraties kwijt kon en zorgden ervoor dat het serieuze werk afgewisseld werd met gewoon gezellig ouwehoeren tijdens de dagelijkse koffiepauzes. Van deze gezelligheid, waaraan ook de vele afstudeerders van de vakgroep een belangrijke bijdrage hebben geleverd, heb ik altijd genoten. Ook, waren deze collega's nooit te beroerd, om iemand eens stevig aan te pakken, om diep te gaan of om een gat te dichten als we met ons zaalvoetbalteam weer een ander team van de mat veegde. Met ons nieuwe kloffie spelen we niet alleen als een sterren ensemble, maar zien we er ook zo uit! Nu alleen nog hopen dat ze eindelijk eens de stand gaan bijhouden zodat we officieel kampioen kunnen worden.

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curriculum vitae

Edwin van Asseldonk werd geboren op 28 maart 1978 te Erp. Hij behaalde in 1996 zijn VWO diploma aan het Zwijsen College te Veghel. Daarna studeerde hij Bewegingswetenschappen aan de Vrije Universiteit van Amsterdam met als differentiatie "Bewegingssysteem". Tijdens zijn afstudeerstage onderzocht hij het effect van een hakverhoging op krachten in de achillespees bij het TNO loop- en schoenencentrum in Eindhoven. Voor deze stage en zijn studieresultaten kreeg hij in 2000 de Gerrit-Jan van Ingen-Schenau beurs uitgereikt. Deze beurs bood hem de mogelijkheid een extra onderzoeksstage uit te voeren in Boston bij het Neuromuscular Research Center van Boston University, waar hij het effect onderzocht van lineaire krachtenvelden op de reikbewegingen. In augustus 2001 rondde hij zijn studie Bewegingswetenschappen cum laude af. Direct na zijn afstuderen werkte hij een periode als onderzoeker en Matlab programmeur bij de afdeling Kinderfysiotherapie van het UMC St. Radboud. In Juni 2002 begon hij als AIO op het LOPES project bij de vakgroep Biomedische werktuigbouwkunde van Universiteit Twente. Tijdens zijn promotie werd hij begeleid door prof. dr. Frans van der Helm en dr. ir. Herman van der Kooij. Als onderdeel van zijn promotie, werkte hij in de zomer van 2005 bij het Neuroimaging and motor control laboratory van Northwestern University in Chicago onder begeleiding van dr. Jules Dewald aan het kwantificeren van de koppelingen in de gewrichtsbewegingen in de benen van CVA patiënten. Dit proefschrift is het resultaat van zijn promotieonderzoek. Vanaf juni 2007 heeft hij zijn werkzaamheden voortgezet bij de vakgroep Biomedische werktuigbouwkunde als universitair docent. In de komende jaren zal hij zich bezighouden met het ontwikkelen en evalueren van trainingen voor CVA patiënten waarbij gebruikgemaakt wordt van revalidatierobots.

list of publications

peer-reviewed journal papers

Nijhuis-Van der Sanden MWG, Van Asseldonk EHF, Eling P & Van Galen GP. (2003). Slow motor performance in girls with Turner Syndrome is not related to increased neuromotor noise. *Motor Control* 7: 111-133

Nijhuis-Van der Sanden MW, Eling PA, Van Asseldonk EH & Van Galen GP. (2004). Decreased movement speed in girls with turner syndrome: a problem in motor planning or muscle initiation? *Journal of Clinical and Experimental Neuropsychology* 26: 795-816

Van Asseldonk EH, Buurke JH, Bloem BR, Renzenbrink GJ, Nene AV, van der Helm FC & van der Kooij H. (2006). Disentangling the contribution of the paretic and non-paretic ankle to balance control in stroke patients. *Experimental Neurology* 201: 441-451

Van Asseldonk EHF, Carpenter MG, Van der Helm FCT & Van der Kooij H. (2007). Use of Induced Acceleration to Quantify the (De)stabilization Effect of External and Internal Forces on Postural Responses. *IEEE Transactions on Biomedical Engineering* 54: 2284 - 2295

Van Asseldonk EHF, Veneman JF, Ekkelenkamp R, Buurke JH, Van der Helm FCT & Van der Kooij H. The effects on kinematics and muscle activity of walking in a robotic gait trainer during zero-force control. *IEEE Transactions on Neural Systems and Rehabilitation Engineering.* Accepted.

Van der Kooij H, Van Asseldonk E & Van der Helm FC. (2005). Comparison of different methods to identify and quantify balance control. *Journal of Neuroscience Methods* 145: 175-203

Van der Kooij H, Van Asseldonk EH, Geelen J, Van Vugt JP & Bloem BR. (2007). Detecting asymmetries in balance control with system identification: first experimental results from Parkinson patients. *Journal of Neural Transmission* 114: 1333-1337

Veneman JF, Kruidhof R, Hekman EEG, Ekkelenkamp R, Van Asseldonk EHF & Van der Kooij H. (2007). Design and Evaluation of the LOPES Exoskeleton Robot for Interactive Gait Rehabilitation. *IEEE Transactions on Neural Systems and Rehabilitation Engineering* 15: 379-386

Veneman JF, Menger J, Van Asseldonk EHF, Van der Helm FCT & Van der Kooij H. Fixating the pelvis in the horizontal plane affects gait characteristics. *Gait & Posture* (In Press). Doi:10.1016/j.gaitpost.2007.11.008

peer-reviewed conference papers

Van Asseldonk EHF, Ekkelenkamp R, Veneman JF, van der Helm FCT & van der Kooij H. (2007). Selective control of a subtask of walking in a robotic gait trainer (LOPES). In *Proceedings of ICORR 2007 - IEEE International Conference on Rehabilitation Robotics* pp. 841-848. Noordwijk, The Netherlands.

Van der Kooij H, Van Asseldonk EHF & Nederhand M. (2007). Detecting asymmetries in balance control with system identification: first experimental results from above knee amputees. In *Proceedings of ICORR 2007 - IEEE International Conference on Rehabilitation Robotics*, pp. 1055-1062. Noordwijk, The Netherlands.

Veneman JF, Van Asseldonk EHF, Ekkelenkamp R, van der Helm FCT & van der Kooij H. (2007). Evaluation of the effect on walking of balance-related degrees of freedom in a robotic gait training device. In *Proceedings of ICORR 2007 - IEEE International Conference on Rehabilitation Robotics* pp. 868-875. Noordwijk, The Netherlands.

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