STEP BY STEP

Stepping strategies to prevent falling while walking.

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CHAPTER 1

General introduction
1.1 Introduction

During walking only one foot is in contact with the ground most of the time. As a result, walking requires a high degree of balance control. Although young, healthy people seem well capable to cope with this challenge, this is not always true for people with gait impairments, for example, due to a lower limb amputation or a hemiparesis due to stroke. Although the causes of gait impairments differ between patient groups, falling appears to be a common problem for many. For example, Miller et al. reported at least one fall in 52.4 percent of lower limb amputees, within the year preceding the study. In a recent review article, Weerdesteyn et al. reported fall rates of 3.8 to 22 percent in the acute phase after stroke, 10.5 to 47 percent during the inpatient rehabilitation phase, and 43 to 70 percent in the chronic phase. Most of these falls occurred during walking. Besides the physical consequences of such a fall, falling also often leads to the development, or an increase, of the fear of falling. This, in turn, limits people in further improving their walking ability and might result in reduced participation in daily life activities. Therefore, a better understanding of the reasons why people with gait impairments fall more often than able-bodied people, and especially about the possibilities to decrease this risk of falling by using specific training methods during rehabilitation is of major importance. The studies reported in this thesis are intended to contribute to this aim by investigating which adaptations of the gait pattern may help to minimize the risk of falling, and whether people with gait impairments can and do employ such strategies. In this general introduction, pertinent literature will be reviewed on how the risk of falling might be quantified and on the possible reasons why people with gait impairments are at an increased risk of falling.

Within this thesis two criteria were defined that have to be fulfilled to prevent falling during daily life walking. First, one has to be stable during steady state walking. Quantifying gait stability is a popular topic in literature, and there are many different measures used for this purpose. In paragraph 1.2.1 the gait stability measures used in this thesis will be motivated and described. The second requirement to prevent falling during daily life walking is the capability to adapt the (stable) gait pattern to a changing environment, in this thesis called gait adaptability. For example, one may need to avoid suddenly appearing obstacles, or maneuver through a busy crowd. In paragraph 1.2.2 a short overview will be given of the studies that attempted to assess gait adaptability. In paragraph 1.3 the studies will be described that have compared people with gait impairments with able-bodied controls in terms of gait stability during steady state walking, and in terms of gait adaptability. The fact that people with gait impairments often differ in their gait pattern, compared to able-bodied people, might be an explanation about why these people are less stable and have less
adaptable gait pattern. An overview of these differences in the gait pattern, and a discussion on how these differences might be related to the increased risk of falling, will be given in paragraph 1.4. In paragraphs 1.5 and 1.6, two set-ups will be described that can be used to experimentally establish whether adaptations in the gait pattern might minimize the risk of falling. In paragraph 1.5 an overview will be given of the studies in which manipulations of the gait pattern were imposed to investigate whether these manipulations influenced gait stability. In paragraph 1.6, studies will be described in which manipulations of gait stability and gait adaptability were used to investigate how subjects adapt their gait pattern in response to these manipulations. Finally, this introduction will be finished with the aim of this thesis and a brief description of the separate chapters (paragraphs 1.7 and 1.8).

1.2. Gait stability and adaptability

In this paragraph an overview will be given of the methods used to assess gait stability and gait adaptability.

1.2.1 Gait stability

To quantify gait stability different measures have been used that highlight different aspects of gait stability and might therefore be differentially related to the risk of falling during walking\textsuperscript{17, 50}. Bruijn\textsuperscript{14} already distinguished performance from robustness. In this context, the performance is the ability of a person to return rapidly to the required operating range after a perturbation, while the robustness reflects the size of a perturbation a person can accommodate without falling. In this thesis the maximum finite-time Lyapunov exponent ($\lambda$) was used as a performance measure, while the margins of stability (MoS) were used to quantify the robustness against a perturbation. $\lambda$, a measure for local dynamic stability, can be described as the capacity to withstand small perturbations that occur naturally during walking\textsuperscript{13, 17, 28-29, 37, 112}. A more exact definition will be provided below. Roughly, $\lambda$ quantifies the rate of divergence from the equilibrium in response to a perturbation. The MoS can be seen as a measure that quantifies the amount of divergence a system can handle before an actual loss of balance occurs. Therefore, high local dynamic stability does not automatically imply a low risk of falling, because the MoS can be very small, and conversely, large MoS might not be large enough in the case of a very low local dynamic stability (figure 1.1). In this paragraph, both stability measures will be explained in more detail.
Figure 1.1: A schematic representation of how the local dynamic stability (LDS), quantified as \( \lambda \), and the margins of stability (MoS) are related to each other. In figures C and D the local dynamic stability is low, a perturbation will result in a fast divergence of the red ball from the equilibrium in contrast to the situations presented in figures A and B. In figures B and D the MoS are large, the divergence from the equilibrium the system can handle before the red ball will leave the bowl is large, in contrast to the situations presented in figures A and C. In the case the ball will leave the bowl, a change of state will take place, which might be a fall in the case of walking.

The maximum finite-time Lyapunov exponent

In recent years, \( \lambda \) has become a more and more popular measure to quantify gait stability\(^{12-13, 15, 17-18, 28, 30-32, 37, 40, 45, 69-71, 77-79, 86, 89, 116, 119-120} \). \( \lambda \) quantifies the average logarithmic rate of divergence of a system after a small perturbation (figure 1.2, upper panel), and therefore an increase in \( \lambda \) implies a decrease in local dynamic stability. A distinction can be made between the short-term Lyapunov exponent (\( \lambda_s \)) and the long-term Lyapunov exponent (\( \lambda_l \)), in which \( \lambda_s \) quantifies the response for a period from 0 to 0.5 (or 0 to 1) stride after a perturbation, while \( \lambda_l \) is often calculated for a longer period after a perturbation, usually 4-10 strides (figure 1.2, lower panel). As discussed above \( \lambda \) is a measure for the local dynamic stability and therefore it might be questionable whether values for \( \lambda \) can be extrapolated to responses to larger perturbations during walking that have the potential to cause a fall. Nevertheless, several studies have shown that \( \lambda \) does reflect balance threats, like perturbations of the walking surface\(^{86} \) and galvanic vestibular stimulation\(^{119} \), which will be discussed further in
paragraph 1.6.1. Besides, it appeared possible to distinguish so-called fallers from non-fallers, by measuring $\lambda$ during steady state walking\textsuperscript{78-79, 116}, which will be discussed in more detail in paragraph 1.3.1.

![Diagram of Lyapunov exponent](image)

**Figure 1.2:** The calculation of the Lyapunov exponent. Upper panel: An example of a 3-dimensional state space and a close-up view of a part of this state space. For each point of the state space, the nearest neighbor is searched, and divergence from these points will be calculated ($s(i)$ versus $s(0)$). Lower panel: The divergence curve, which expresses the logarithmic rate of divergence, from which maximum time finite Lyapunov exponents, $\lambda_s$-stride and $\lambda_l$-stride can be calculated as the slope of the curve at 0–0.5 strides and at 4–10 strides, respectively.

**Margins of stability**

Traditionally, balance maintenance while standing is believed to require that the projection of the centre of mass (CoM) has to fall within the margins of the base of support (BoS). However, Pai and Patton\textsuperscript{98} among others have shown that this condition alone is not sufficient to maintain balance during standing, because the horizontal velocity of the CoM
affects balance and a more dynamic concept is thus needed. This can be achieved by assessing motion state of the CoM (i.e. its position and velocity) in relation to the BoS. In the study of Pai and Patton\textsuperscript{98} a simple inverted pendulum model with one foot, which provides a BoS, was used to define a, so-called, feasibility region existing of the combinations of the position and velocity of the CoM with respect to the BoS that will not result in a forward or backward fall due to a loss of balance (figure 1.3). The same concept can be used to quantify stability during walking. Hof et al.\textsuperscript{57} provided an analytical expression for this model and extended it to the medio-lateral direction. They calculated the MoS as the difference between the position of the centre of pressure (CoP; as the margin of the BoS) and the extrapolated centre of mass (XCoM). The XCoM is defined as the position of the CoM, plus its velocity times a factor $\sqrt{l/g}$, with $l$ being the length of the pendulum (for which often the leg length is used), and $g$ the acceleration due to gravity\textsuperscript{57}.

\textbf{Figure 1.3:} Feasible horizontal centre of mass velocity-position (shaded diagonal band) for terminating anterior movement of a simple pendulum connected to a stationary base of support, calculated by the model of Pai and Patton\textsuperscript{98}, derived from their Fig. 3.
In the studies presented in this thesis, with the exception of chapter 2 where we followed Hof’s definition more closely, we calculated the MoS in medio-lateral (ML; figure 1.4a) and backward direction (BW; figure 1.4b) direction, as respectively the minimum distance between the XCoM and the lateral border of the standing foot and the distance between the XCoM and the backward border of the leading foot at initial contact. A negative ML MoS, implying that the XCoM is located lateral of the border of the BoS, will result in a deviation from the straight walking trajectory. A negative BW MoS, implying that the XCoM is located posterior to the border of the BoS of the leading foot, will result in an interruption of the forward progression. Both these situations can be considered as unstable, because in case of a negative MoS, a recovery response, in the form of a crossing step or a backward step will be necessary to prevent, respectively, a sideward or backward fall. An increase in ML MoS and BW MoS will make people more robust against perturbations that have the potential to cause a sideward or backward loss of balance during the single support phase. It has to be noticed that an increase in BW MoS has the disadvantage that the risk of a forward loss of balance will increase. However, in contrast to a backward loss of balance, in this case a (small) adaptation of the next step(s), like a temporal increase in step length, might be sufficient to regain balance.

1 Note that for the BW MoS our definition of a positive MoS is opposite to that of Hof et al. which we did follow in chapter 2. Our definition of a positive BW MoS, is based on the equation XCoM-BoS, instead of the original equation BoS-XCoM, such that maintenance of forward progression, one of our definitions of a stable gait pattern, is reflected in a positive BW MoS.
1.2.2 Gait adaptability

In contrast to the assessment of gait stability, there are no specific outcome measures that can be used to quantify gait adaptability. Therefore, the evaluation of gait adaptability strongly depends on the experimental set-up that has been used. In this paragraph, an overview will be given of these set-ups and of how these set-ups contribute to the quantification of gait adaptability.

Multiple studies have used obstacle avoidance tasks to assess gait adaptability. Within these studies a distinction can be made between experiments that were performed on a walkway where obstacles were visible several steps before the obstacle had to be crossed and experiments that were performed on a treadmill with a set-up that allowed to present an obstacle.

Figure 1.4: Schematic representation of the definition of the backward (BW) and medio-lateral (ML) margin of stability (MoS). A: The ML MoS is defined as the minimum distance in medio-lateral direction between the extrapolated centre of mass (XCoM; dotted line in the right panel) and the lateral border of the foot during foot-contact. The XCoM is calculated as the position of the centre of mass (CoM; solid line in the right panel) plus its velocity (vCoM) times a factor \( \sqrt{\frac{l}{g}} \), with \( l \) being the length of the pendulum (for which often the leg length is used) and \( g \) the acceleration of gravity. B: The BW MoS is defined as the distance in anterio-posterior (AP) direction between the XCoM and the posterior border of the leading foot (dashed line in the right panel) at initial contact (IC).
obstacle just before the moment the obstacle had to be crossed\textsuperscript{27,60,65}. The latter experiments mimic situations in which sudden and fast adaptations of the gait pattern are necessary. In both experimental set-ups, the amount of obstacle hits was used to quantify how well subjects are able to adapt their gait pattern to the environment.

More recently virtual targets were used to assess gait adaptability. An advantage of virtual targets, instead of a physical obstacle, is the possibility to use a more continuous outcome measure to quantify the performance on the obstacle avoidance task\textsuperscript{65}. Instead of the dichotomous outcome ‘hit’ or ‘no hit’, it is possible to take, for example, the amount of overlap between foot and obstacle to quantify the success rate, which can, in turn, be used as an outcome measure for gait adaptability.

During daily life walking, maintaining stability and facilitating gait adaptations cannot be separated from each other. While crossing an obstacle, not only the accuracy of the adaptation to prevent an obstacle hit is of importance, but also the maintenance of balance while performing this task. Only a single study addressed the coupling between stability and adaptability previously\textsuperscript{108}, and for the stride in which the obstacle had to be crossed. Therefore, this aspect is still underexposed.

1.3 Gait stability and adaptability: people with gait impairments versus able-bodied controls

In this paragraph, an overview will be given of the studies that compared people with gait impairments with able-bodied controls in terms of gait stability during steady state walking, and gait adaptability.

1.3.1 Gait stability

With respect to $\lambda$, several studies have compared elderly with and without a history of falling, and found higher values for $\lambda$s for the so-called fallers compared to the non-fallers\textsuperscript{78-79,116}, while the long-term Lyapunov exponent $\lambda_l$ appeared not to be different between groups\textsuperscript{116}. Also for transfemoral amputees higher values for $\lambda$s were found compared to able-bodied controls\textsuperscript{77}. These results provide evidence for the face validity of $\lambda$s as a potential indicator of a higher fall risk of people with gait impairments, while the $\lambda_l$ appeared not to provide information about fall risk\textsuperscript{116}. The weak correlation between $\lambda_l$ and the actual risk of falling is also supported by the results of modeling studies\textsuperscript{16,103} and therefore in this thesis, local dynamic stability will be assessed only by means of the $\lambda$s.

With respect to the MoS, there are a limited number of studies regarding people with gait impairments. The available studies compared amputees (transtibial\textsuperscript{23}, transfemoral\textsuperscript{58}, and
bilateral\textsuperscript{81}) with able-bodied controls with respect to their ML MoS. Amputees walked with larger ML MoS, despite a larger body sway in the frontal plane\textsuperscript{81}. Besides, for the unilateral amputees it appeared that the ML MoS at the side of the amputation was larger compared to the healthy side\textsuperscript{23,58}. The larger ML MoS for amputees was realized by a larger step width for amputees compared to the able-bodied controls. The reason why amputees walk with a larger ML MoS is probably because they are not able to use the so-called ‘ankle strategy’ to correct the ML MoS after foot placement. Therefore it is assumed that amputees create a safety margin that is large enough to recover from a potential perturbation.

1.3.2. Gait adaptability

As already mentioned in paragraph 1.2.2, for assessment of gait adaptability, a distinction can be made between experiments in which obstacles were already visible several steps before the obstacle had to be crossed\textsuperscript{54-55,65,107-108} and experiments in which obstacles were presented just before the moment the obstacle had to be crossed\textsuperscript{27,60,65}. For people with gait impairments the latter situation appeared to be far more challenging. For example, amputees\textsuperscript{60,65} and post-stroke individuals\textsuperscript{27}, hit obstacles more often during crossing compared to able-bodied controls, especially when the available response time becomes shorter\textsuperscript{27,60}. In addition, the study of Said et al.\textsuperscript{110} indicated that post-stroke individuals also have an impaired postural balance during obstacle crossing, which might further increase fall risk.

1.4 Gait pattern differences between people with gait impairments and able-bodied people

People with gait impairments often differ in their gait pattern compared to able-bodied controls. Several studies have compared spatio-temporal parameters between these groups, and have tried to relate these differences in gait pattern to the risk of falling. People with gait impairments often walk slower with shorter steps, at a lower step frequency and with a larger step width compared to able-bodied controls\textsuperscript{23,44,58,74,99,122,124}. However, it is still under debate whether and how these differences in the gait pattern might be related to the risk of falling during walking. Some investigators have concluded that this deviant gait pattern might be a predictor of future falls\textsuperscript{40}, while others concluded that there is no relation between these differences in the gait pattern and the chance of falling during walking\textsuperscript{51-52}. In line with the latter, Maki\textsuperscript{82} found that a slower walking speed is related to the fear of falling, but not to the actual risk of falling. In contrast, Winter et al.\textsuperscript{128} concluded that walking slower with a shorter step length reflects a ‘more cautious’ gait strategy and might decrease the risk
of falling. With respect to step width, the opinion about the relationship with falling seems to be more unanimous, namely that a larger step width will increase the size of the base of support, and therefore reflects a positive adaptation to decrease the risk of falling\textsuperscript{19,35}.

Besides these overall gait differences between people with and without gait impairments, unilateral impairments often cause an asymmetry between legs during walking. Although such an asymmetry is often seen as a detrimental effect of the impairment, it might be a functional compensation as well. For example, the results of studies performed by Houdijk et al.\textsuperscript{64} and Kuo et al.\textsuperscript{33} suggest that for amputees, decreasing step length when pushing off with the prosthetic leg might be a strategy to limit the higher metabolic cost during the step to step transition, caused by the lack of ankle push-off. The link between step asymmetry and the risk of falling has never been made before, but it might be of interest to evaluate the step asymmetry in view of gait stability.

One of the problems in literature relating differences in gait pattern with gait stability resides in the design of these studies. It is not inconceivable that the observed differences in gait pattern also serve other purposes, like, for example, optimizing the energetic demands of walking. Manipulating the gait pattern while measuring the effect of these manipulation on, for example, local dynamic stability and margins of stability, or monitoring gait adaptations in response to manipulations with the intention to increase the risk of falling, might give us more insights into the question whether gait pattern deviations of people with gait impairments might contribute to the minimization of the risk of falling. These methods will be described further in paragraph 1.5 and 1.6.

1.5. Gait pattern and gait stability

To gain a better understanding of how differences in gait pattern between individuals with gait impairments and able-bodied people might be related to fall risk, several studies have investigated whether imposed adaptations of the gait pattern influence gait stability. In the next section an overview will be given of the studies that related different aspects of the gait pattern to respectively \(\lambda_s\) and the MoS.

\textit{Short-term Lyapunov exponent}

Most studies that have related the gait pattern to the \(\lambda_s\) investigated the effect of differences in walking speed\textsuperscript{13,15,18,31,37,70,83,112}. However, the results of these studies are inconsistent. Although multiple studies have found an increase in \(\lambda_s\), and therefore a decrease in local dynamic stability, with an increase in walking speed\textsuperscript{29,31,70}, there are also studies in which the relationship between walking speed and \(\lambda_s\) was less clear\textsuperscript{13,15,83,112}, or in which a decrease in \(\lambda_s\) was found as result of an increase in walking speed\textsuperscript{18}. An explanation for these
differences in results might be the lack of a uniform method to calculate $\lambda s$. The studies mentioned above used different sources of kinematic input (for example trunk velocity or acceleration, joint angles, orientations of lower leg and thigh, and angular velocity or acceleration of the trunk) and different methods to reconstruct state spaces (different numbers of embedding dimensions and different methods to increase the embedding dimensions, i.e. using time-delay copies or using different sources of kinematic input). Besides, different methods were used to equalize the length of the signals between trials (a constant amount of gait cycles, a constant amount of samples or a constant amount of gait cycles after time-normalizing the time-series). Finally, different time frames were used over which $\lambda s$ was calculated (from 0-1 stride or from 0-0.5 stride). Another explanation for the differences in results of the studies described above might be the uncontrolled effect of the different combinations of stride length and stride frequency that are associated with walking at different walking speeds. Therefore, McAndrew and Dingwell investigated the effect of different stride lengths on $\lambda s$, and they found clear effects in this regard. This result contributes to the presumption that a possible effect of walking speed on $\lambda s$ might be influenced by the underlying ratio of stride length and stride frequency.

**Medio-lateral and backward margins of stability**

In contrast to the calculation of $\lambda s$, the underlying mechanical model used to calculate the ML and BW MoS yields direct predictions on the effect of different aspects of the gait pattern on the size of the ML and BW MoS. For both the size of the ML and BW MoS the placement of the foot is an important mechanism to regulate the size of the MoS. For ML MoS, an increased step width results in the placement of the edge of the BoS more lateral with respect to the extrapolated centre of mass (XCoM) and will therefore increase ML MoS. An increase in BW MoS can be attained by placing the leading foot more backward with respect to the CoM, which can be realized by decreasing step length. Besides the foot placement, MoS can be increased by influencing the trajectory of the XCoM. An increase in step frequency will result in a smaller excursion of the XCoM in sideward direction, which will increase ML MoS. An increase in walking speed will increase the forward velocity of the CoM, which will bring the XCoM further forward with respect to the leading foot, and will therefore increase BW MoS. Espy et al. tested the effects of walking speed and step length on the BW MoS and indeed found an increase in BW MoS when subjects walked faster or with shorter steps compared to their comfortable gait pattern. McAndrew and Dingwell tested the effect of different step lengths and step widths on the size of the ML and BW MoS. They found an increase in ML MoS when they asked subjects to walk with wide steps and an increase in BW MoS when they
asked subjects to walk with short steps, which was also in line with the underlying mechanical model. However, because of the experimental set-up of these studies, subjects walked at a fixed walking speed, the step length manipulation automatically implied a step frequency manipulation. In addition, during the step width manipulation in the study of McAndrew and Dingwell, step length and step frequency were not controlled. Both limitations might have biased the results. To investigate the effect of walking speed, step length, and step frequency on gait stability independently, a protocol in which these parameters are manipulated systematically will be necessary. Such an experimental set-up was already used before by Bertram and Danion et al. to examine the effect of these parameters on energy cost and gait variability.

1.6. Manipulations of gait stability and gait adaptability

To investigate whether adaptations of the gait pattern are functional to increase or at least maintain gait stability and/or gait adaptability, these aspects of walking should be challenged. Manipulations of gait stability and adaptability have been used previously to investigate whether able-bodied subjects, but also amputees and post-stroke individuals, are capable of adapting their gait pattern to withstand these manipulations.

1.6.1. Manipulations of gait stability

A possible method to manipulate gait stability is the use of continuous balance perturbations during walking. Continuous perturbations that were used before are galvanic stimulation, virtual environments with complex scenes, tilted scenes, oscillating scenes, and translating walking surfaces. From studies of Van Schooten et al. and McAndrew et al. it appeared that the perturbations caused an increase in stability, and therefore a decrease in local dynamic stability, but an increase in ML and BW MoS. The increase in ML and BW MoS coincided with a decrease in step length and an increase in step width in response to the perturbation. These results might imply that subjects became locally less stable in response to the perturbations, but were capable to increase their margins of stability to prevent an actual fall.

In all studies described in the previous subparagraph, subjects walked on a treadmill at a fixed walking speed and were therefore not able to adapt walking speed to the perturbations. This is an important limitation, because an adaptation in walking speed might be functional in the regulation of gait stability. Besides, all the studies described above were performed in able-bodied controls. Therefore, the results of these studies might provide indications of strategies used to maintain gait stability in a challenging walking environment, but it is
unknown whether people with an impaired gait are able to use these strategies, to the same
degree, as able-bodied people.

1.6.2. Manipulations of gait adaptability
In paragraph 1.2.2, a description of the experimental set-ups in which (physical and virtual)
obstacles were used to assess gait adaptability was already given, while in paragraph 1.3.2
the differences in performance on these tasks between people with gait impairments and
able-bodied controls were described. In addition, some of these studies have also looked at
the strategies used by the different subject groups to perform an obstacle avoidance task. It
appeared for example that both amputees\textsuperscript{60} and post-stroke individuals\textsuperscript{27} preferred
lengthening of the crossing step, while able-bodied controls preferred shortening of the pre-
crossing step. This difference in strategy might be a cause of the larger failure rate for both
patient groups compared to the able-bodied controls.
The studies described in the previous subparagraph measured only differences in the
response on the appearance of the obstacle, and do not describe whether subjects adapted
their strides before obstacle crossing as a anticipatory strategy. Such anticipatory strategies
might also be used during daily life walking when people are aware of potential changes in
the environment. The use of anticipatory strategies might, first, serve the purpose to
facilitate the fast and accurate adaptation necessary to prevent an obstacle hit. However,
secondly a loss of balance as a result of the fast adaptation has to be prevented, which
appears to be, among others a problem in post-stroke individuals\textsuperscript{108}. Therefore, anticipatory
strategies might also be used by individuals with gait impairments, but also by able-bodied
controls, to preserve gait stability during the adaptation.

1.7 General aim and research questions
Based on the literature covered in this general introduction, we can conclude that there is
still a lot unknown about how gait adjustments can contribute to a decrease in fall risk.
Besides, the ability of able-bodied people, but especially people with gait impairments to use
such adjustments as a strategy to decrease fall risk in challenging walking conditions need to
be further investigated. Therefore, the general \textbf{aim} of the studies presented in this thesis was
\textbf{to investigate which gait strategies can be used to decrease fall risk during walking and
whether people with gait impairments are able to use these strategies}. Knowledge of
these strategies might be input for protocols to evaluate walking skills, in terms of
minimizing fall risk, during a rehabilitation trajectory.
To investigate whether strategies that can and will be used to minimize fall risk during
walking are generic for different gait impairments, besides a group of able-bodied controls,
two groups of subjects with different gait impairments were included in the studies presented in this thesis. Firstly, a group of subjects with a gait impairment caused by a peripheral disorder, namely a transtibial amputation, and secondly, a group of subjects with a gait impairment caused by a central affection, namely stroke.

The research presented in this thesis aims to answer the following specific research questions:

1. Which gait strategies do able-bodied people use to cope with manipulations of gait stability and gait adaptability?
2. How do manipulations of stride frequency, stride length and walking speed affect the short-term Lyapunov exponent and the margins of stability in able-bodied people?
3. Which gait strategies do transtibial amputees and post-stroke individuals use to withstand manipulations of gait stability and gait adaptability and do these strategies differ from the strategies used by able-bodied controls?
4. How does step length asymmetry in amputees contribute to the regulation of the backwards margins of stability?

1.8 Outline of the thesis

Which gait strategies do able-bodied people use to cope with manipulations of gait stability and gait adaptability?

To investigate which strategies people use, in terms of adjusting walking speed, step frequency, step length, and step width, to withstand a manipulation of gait stability, a continuous sideward translation of the walking surface was used in chapter 2. To measure whether gait stability changed in response to this perturbation, we calculated $\lambda$, a measure for the local dynamic stability and the MoS in ML and BW direction. Besides being stable during steady state walking, to prevent falling, one has to be able to make (fast) adjustments of this stable gait pattern to environmental constraints, for example to avoid an obstacle. In chapter 3 we will describe whether able-bodied subjects adjust walking speed, step frequency, step length, and step width in response to a gait adaptability task (GA-task) in which healthy subjects had to hit virtual targets that sudden appeared on the screen with their knees. To quantify gait stability, while performing the GA-task, ML and BW MoS were calculated.
How do manipulations of stride frequency, stride length and walking speed affect the short-term Lyapunov exponent and the margins of stability in able-bodied people?

To answer this research question, in chapter 4 we systematically manipulated step length and step frequency, and consequently walking speed, in a group of healthy controls. With this experimental set-up it is possible to investigate how adjustments in these gait parameters independently affect the size of the ML and BW MoS and the $\lambda$s. The results of this study might provide an additional explanation for the gait adjustments found in response to manipulations of gait stability and adaptability (chapter 2 and 3).

Which gait strategies do transtibial amputees and post-stroke individuals use to withstand manipulations of gait stability and gait adaptability and do these strategies differ from the strategies used by able-bodied controls?

The results of the studies described in chapter 2, 3 and 4 help to create a reference with respect to the strategies that can be used to increase or at least maintain gait stability during challenging walking conditions, in the absence of any gait impairments. In these chapters the strategies used by transtibial amputees (chapter 5) and post-stroke individuals (chapter 6) to enhance ML and BW MoS during manipulations of gait stability and gait adaptability will be described, and how these strategies differ from strategies used by able-bodied controls. Besides we measured $\lambda$s to investigate whether local dynamic stability differed between amputees and able-bodied controls and whether local dynamic stability was affected in these groups as a consequence of the manipulation of gait stability (chapter 5).

How does step length asymmetry in amputees contribute to the regulation of the backwards margins of stability?

In the preceding chapters, we focused on the role of average adjustments in step parameters over both legs. However, one of the characteristics of people with unilateral gait impairment is an asymmetry between legs during walking. In chapter 7, an observational study is described in which we have investigated whether the step length asymmetry in amputees may be functional with respect to the regulation of the size of the BW MoS.

In the chapter 8 the results and conclusions of the studies described in the previous chapter will be summarized and discussed. Furthermore, implications for the rehabilitation practice and clues for further research will be presented.
CHAPTER 2

Speeding up or slowing down? Gait adaptations to preserve gait stability in response to balance perturbations

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2.1 Abstract

It has frequently been proposed that lowering walking speed is a strategy to enhance gait stability and to decrease the probability of falling. However, previous studies have not been able to establish a clear relation between walking speed and gait stability. We investigated whether people do indeed lower walking speed when gait stability is challenged, and whether this reduces the probability of falling.

Nine healthy subjects walked on the Computer Assisted Rehabilitation ENvironment (CAREN) system, while quasi-random medio-lateral translations of the walking surface were imposed at four different intensities. A self-paced treadmill setting allowed subjects to regulate their walking speed throughout the trials. Walking speed, step length, step frequency, step width, local dynamic stability (LDS), and margins of stability (MoS) were measured.

Subjects did not change walking speed in response to the balance perturbations (p=0.118), but made shorter, faster, and wider steps (p<0.01) with increasing perturbation intensity. Subjects became locally less stable in response to the perturbations (p<0.01), but increased their MoS in ML (p<0.01) and backward (p<0.01) direction.

In conclusion, not a lower walking speed, but a combination of decreased step length and increased step frequency and step width seems to be the strategy of choice to cope with medio-lateral balance perturbations, which increases MoS and thus decreases the risk of falling.

Key words
Balance perturbations; Walking speed; Step length; Step frequency; Step width
2.2 Introduction

It is generally assumed that walking speed is reduced to cope with an increased probability of falling, due to environmental or internal disturbances of stability\(^{31, 37, 70}\). In line with this assumption, several recent studies have examined the effects of walking speed on local dynamic stability (LDS)\(^{13, 31, 37, 70}\). LDS, as captured by the short-term Lyapunov exponent (\(\lambda_s\)) reflects the response of people to small perturbations, and is therefore seen as an index of the stability of human walking\(^{31}\). Besides LDS has been shown to be correlated to balance impairments and fall risk\(^{77-79, 116, 119}\). However, studies on the relation between walking speed and LDS have provided contradictory and inconclusive results\(^{13, 31, 37, 70}\). It therefore remains unclear whether slow walking is more stable than fast walking.

Besides using LDS-measures, the probability of falling has been analyzed in terms of margins of stability (MoS)\(^{38, 57-58, 98}\). This measure has a more logical relationship with the probability of falling than LDS-measures. The MoS is quantified as the distance between the centre of mass (CoM) motion state (i.e. its position and velocity) relative to the base of support. Mathematical modelling of human walking as an inverted pendulum allows for theoretical hypotheses on the effect of walking speed on MoS and therefore on the probability of falling\(^{98}\). Following this model, the probability of making a backward fall can be reduced by decreasing step length or increasing CoM velocity, the latter being directly related to an increased walking speed\(^{38, 98}\). Hof et al.\(^{57, 59}\) provided an analytical expression for this model and extended it towards the probability of falling in ML-direction\(^{58}\). This model does not predict a direct influence of walking speed on the size of the MoS in ML-direction, but it does predict that the ML MoS will increase by increasing step width and step frequency\(^{58}\), the latter could coincide with an increase in walking speed.

Hence, despite the common belief that reducing walking speed might reduce the probability of falling; there is little empirical evidence to support this. Moreover, although several studies have attempted to assess the effect of walking speed on gait stability, it has never been assessed whether people really select a slower walking speed in challenging conditions and how possible changes in walking speed affect LDS and MoS.

The main purpose of this study was to investigate if subjects adapt walking speed when gait stability is challenged. For this purpose we used a self-paced treadmill, which made it possible for subjects to continuously adapt their walking speed to imposed ML balance perturbations. Simultaneously, we observed the effect of the balance perturbations on step length, step frequency, and step width. Subsequently, we assessed the effect of potential adaptations in walking speed on LDS and the MoS in AP- and ML-direction, by making a
comparison between trials at self-paced speed and trials at a fixed speed, in which subjects could not adapt walking speed to the perturbations.

2.3 Methods

2.3.1 Subjects
Nine healthy adult subjects (4 men and 5 women, age: 32.2±7.5 years, height: 1.77±0.12 m, weight 72.2±14.0 kg) were included. The study was approved by the local ethical committee and subjects gave written informed consent prior to their participation.

2.3.2 Equipment
All subjects walked on the Computer Assisted Rehabilitation ENvironment (CAREN) system (Motek Medical b.v., Amsterdam, The Netherlands). The CAREN system consists of an instrumented treadmill mounted onto a 6-degree-of-freedom motion platform in combination with a Virtual Environment (VE) projected on a 180° semi-cylindrical screen (figure 2.1 upper left panel). During the experiment, the motion platform was used to induce perturbations in ML-direction during walking. The VE used in this experiment was a virtual road, surrounded by trees to create an optical flow while walking on the treadmill (figure 2.1 upper right panel). Motek Medicals D-flow software was used to control the system and to synchronize the instantaneous treadmill speed and scene progression. Twelve high resolution infra-red cameras (Vicon, Oxford, UK) and the Vicon Lower Body Plug-in-Gait marker set were used to capture kinematic data. For a part of the experimental trials the treadmill was used in the self-paced mode. This allowed subjects to modify walking speed at will, by servo-controlling the motor with a real time algorithm that took into account the pelvis position in the AP-direction, as measured by the markers attached to the pelvis, and a reference position corresponding to the AP-midline of the treadmill. A safety harness system suspended overhead prevented the subjects from falling but did not provide any weight support.
Figure 2.1: Upper left panel: Experimental setup: CAREN. Upper right panel: Virtual environment used in the experiment. Lower panel: Time-series for one period of the perturbation with a scaling factor (A) of 0.2. Frequency of this perturbation varied from 0.16 to 0.49 Hz, maximum excursion of the platform during this perturbation was 0.67 m.

2.3.3 Protocol

Warming up

Subjects completed two warming-up trials of 3 minutes, to become familiar with walking on the (self-paced) treadmill and the VE, before the protocol started. During the first warming-up trial subjects walked at a fixed walking speed, around their comfortable walking speed. During the second warming-up trial, subjects had the opportunity to practice with the self-paced mode of the treadmill, but were asked to walk at comfortable walking speed during the final minute.

Experimental trials

The protocol consisted of 10 trials of 4 minutes walking. Besides an unperturbed condition, balance perturbations were applied, at four different intensities, by translating the walking surface in ML direction, following a multi-sine function, which made the perturbation
unpredictable for the subjects. This perturbation was already used before by McAndrew et al.\textsuperscript{85-86}.

\[ D(t) = A \left[ 1.0\sin(0.16 \times 2\pi t) + 0.8\sin(0.21 \times 2\pi t) + 1.4\sin(0.24 \times 2\pi t) + 0.5\sin(0.49 \times 2\pi t) \right] \]

where \( D(t) \) is the translation distance (m), \( t \) is time (s), and \( A \) is the scaling factor which was used to vary the intensity of the perturbation. Scaling factors of 0.05, 0.10, 0.15 and 0.20 were used. To illustrate the character of the perturbation the pattern of the perturbation with a scaling factor of 0.2 is shown in the lower panel of figure 2.1. The four perturbation conditions and the unperturbed trial were all repeated twice, once at self-paced walking speed, where subjects continuously had the opportunity to adapt their walking speed to the balance perturbations, and once at a fixed walking speed, where subjects did not have the opportunity to adapt their walking speed to the balance perturbations. Fixed walking speed was set to the comfortable walking speed that was measured during warming up in the self-paced mode, and was therefore comparable with a comfortable walking speed during unperturbed walking. During the self-paced conditions subjects started at a fixed speed of 1 m/s. After 30 s., the self-paced mode was switched on. Before every trial instructions about the upcoming trial were given. All trials were imposed at random.

\textbf{2.3.4 Data collection}

During the trials in which the subjects walked at a self-paced walking speed, the speed of the treadmill was recorded. Kinematic data of markers attached at the lateral malleoli of the ankles and the pelvis (left and right anterior superior iliac spines (LASI & RASI), and left and right posterior superior iliac spines (LPSI & RPSI)) were collected with the Vicon system at a sampling rate of 120 Hz. The last three minutes of each trial were used for data analysis. Before data analysis, both speed data and kinematic data, except for the calculation of LDS, were low-pass filtered with a bi-directional Butterworth filter with a cut-off frequency of 2 Hz.

\textbf{2.3.5 Data analysis}

\textit{Walking speed}

Walking speed was calculated as the average treadmill speed over the last three minutes for every trial executed at self-paced walking speed.

\textit{Step parameters}

Step frequency was determined as the inverse of the average duration between two subsequent heel-strikes, where heel-strikes were detected as the local maxima of the position of the ankle markers in the AP-direction. Step width was calculated as ML-distance
between both ankle markers at the moment of heel-contact and step length was defined as the AP-distance between these markers at the moment of heel-contact.

**Local dynamic stability**

To calculate LDS, position data of the markers placed on LASI, RASI, LPSI, and RPSI were used. Given the difficulties associated with filtering nonlinear signals, data were analyzed without filtering. Local pelvis reference frames were defined following the ‘Conventional Gait Model’ [25, 68]. Linear accelerations of the ML, AP, and vertical (VT) direction were calculated as the second derivative of the position of the origin. Rotational velocities in three directions were calculated following the method defined by Zatsiorsky [130]. The first 150 consecutive strides of each time-series were analyzed, because estimates of LDS may be biased by time-series length and number of strides [12, 69]. Time-series were time-normalized, using a shape-preserving spline interpolation, such that each time-series of 150 strides had a total length of 15,000 samples [13, 37]. Subsequently, 12D state spaces were reconstructed from the time-normalized 3D linear acceleration and 3D rotational velocities time series, each with their 25 samples delayed copies [15, 69-70].

From the constructed state spaces, Euclidean distances between neighbouring trajectories in state space were calculated as a function of time and averaged over all original nearest neighbour pairs to obtain the average logarithmic rate of divergence. The slope of the resulting divergence curves for the interval between 0-50 samples provides an estimate of LDS in terms of the short-term Lyapunov exponent for the period of approximately 0-1 step ($\lambda_{S\text{-step}}$) [13, 104].

**Margins of stability**

To calculate the margins of stability (MoS), a method derived from the procedure used by Hof was used [57-58]. In this study the extrapolated centre of mass (XCoM), was defined as the origin of the local pelvis reference frame [25, 68], representing the CoM, plus its velocity times a factor $\sqrt{l/g}$, with l being the maximal height of the origin of the pelvis and g the acceleration of gravity. The MoS were calculated for both the AP- and ML-direction as the position of the lateral malleolus of the ankle of the leading foot (representing the border of the base of support) minus the position of the XCoM for the moment at which the MoS reached its minimum value within the period of one step [57-58]. Although basically similar our method

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2In contrast to the other chapters of this thesis, in this chapter we used the original mathematical equation BoS-XCoM to determine the MoS in AP direction. Note that a negative AP MoS in this chapter corresponds to a positive BW MoS used in the rest of this thesis.
differs from Hof$^{57-58}$ who used force plate data for calculating XCoM and MoS. MoS was averaged for the first 150 steps of the last three minutes of each trial.

### 2.3.6 Statistical design

2 × 5 within factorial ANOVAs were performed, with perturbation intensity and speed condition (self-paced or fixed speed) as within variables, to search for significant differences in walking speed, step length, step frequency, step width, $\lambda_{S\text{-step}}$, and the MoS in AP- and ML-direction. When a significant effect of perturbation intensity was found, simple contrasts were used to investigate for which particular perturbation intensities the concerning variable differed from unperturbed walking. Significant interaction effects were further investigated by executing paired-samples t-tests with a Bonferroni correction for each perturbation condition. Statistical analyses were performed using SPSS 17.0 (SPSS Inc., Chicago, IL).

### 2.4 Results

All subjects completed the experiment and no one fell during the perturbation trials. During the self-paced trials, subjects did not significantly lower their walking speed in response to the balance perturbations ($F=2.00$; $p=0.118$; $df=4$; figure 2.2). During the fixed speed trials, walking speed was lower compared to the self-paced trials ($F=11.034$; $p=0.011$; $df=1$); however, no interaction with perturbation intensity was found ($F=2.00$; $p=0.118$; $df=4$).

Despite the absence of adaptations in gait speed to the different perturbation intensities, the underlying step parameters, step length and step frequency, were significantly affected by the perturbations. Step length decreased with perturbation intensity ($F=23.118$; $p<0.01$; $df=4$), and differed from unperturbed walking for the second to fourth perturbation intensity (figure 2.3a). Step frequency increased with an increase in perturbation intensity ($F=35.713$; $p<0.01$; $df=4$), and differed from unperturbed walking for all perturbation intensities (figure 2.3b). Step width was significantly larger at all the perturbation intensities, compared to unperturbed walking ($F=26.805$; $p<0.01$; $df=1.457$ after a Greenhouse-Geisser correction for non-sphericity) (figure 2.3c). Step length and step width did not differ between the two speed conditions. Step frequency was higher for the self-paced condition, compared to the fixed speed condition ($F=16.983$; $p<0.01$; $df=1$). No interactions were found for these step parameters.

A significant increase in $\lambda_{S\text{-step}}$ with perturbation intensity ($F=28.090$; $p<0.01$; $df=4$) was observed and $\lambda_{S\text{-step}}$ differed from unperturbed walking from the second perturbation intensity (figure 2.4a). No main effect of speed condition was found on $\lambda_{S\text{-step}}$ ($F=2.460$; $p=0.118$; $df=4$).
p=0.155; df=1), but there was a significant interaction of perturbation intensity and speed condition (F=3.670; p=0.014; df=4). However, a post-hoc test with Bonferroni correction showed a significant difference in $\lambda_S$-step between self-paced and fixed-speed walking only for the unperturbed condition (p=0.015).

MoS were all positive in ML-direction and negative in AP-direction at initial contact which means that the XCoM was located medial and anterior with regards to the marker attached to the lateral malleolus of the leading foot (Figure 2.4b). ML MoS increased (F=8.041; p=0.016; df=1.275 after a Greenhouse-Geisser correction for non-sphericity) and AP MoS decreased, i.e. became more negative (F=24.001; p<0.01; df=4) in response to the perturbations, and differed from unperturbed walking for all perturbation intensities. For the AP MoS a main effect of speed-condition was found (F=7.392; p = 0.03; df=1).

![Figure 2.2: Average and standard deviation of walking speed (n = 9) for all experimental conditions. Black bars represent self-paced walking and white bars represent fixed speed walking (U = unperturbed walking; P1 to P4 = Perturbation intensity 1 to 4).](image-url)
Figure 2.3: Averages and standard deviations of step length (A), step frequency (B), step width (C), (n = 9) for all experimental conditions. Black bars represent self-paced walking and white bars represent fixed speed walking (U = unperturbed walking; P1 to P4 = Perturbation intensity 1 to 4). * indicates the perturbation intensities, for which the concerning step parameter differed from unperturbed walking.
2.5 Discussion

In this study, we investigated whether and how healthy people adapt walking speed when stability is challenged, and how these speed changes affect local dynamic stability, as captured by $\lambda_s$ and margins of stability (MoS) in both AP- and ML-direction. Notably, subjects did not lower their walking speed in response to the balance perturbations, which is in contrast to the assumption that lowering walking speed is a strategy to decrease the probability of falling$^{31, 37, 70}$. However, subjects did change the underlying step parameters, step frequency and step length. Decreases in step length and increases in step frequency were found with increasing perturbation intensity. In addition, subjects increased their step width in response to the perturbations. These changes were also found when walking speed was kept constant across all perturbation conditions, in accordance with previous findings of McAndrew et al.$^{85}$. 

Figure 2.4: Average and standard deviation of $\lambda_s$-step (A) and MoS in ML- (upper panel) and AP-direction (lower panel) (n=9) for all experimental conditions. Black bars represent self-paced walking and white bars represent fixed speed walking (U = unperturbed walking; P1 to P4 = Perturbation intensity 1 to 4). * indicates the perturbation intensities, for which $\lambda_s$-step and the MoS differed from unperturbed walking.
Although walking speed did not change with perturbation intensity a systematic speed difference between the self-paced and fixed speed condition was found. Probably the fixed speed determined during the practice trial was an underestimation of preferred walking speed. Because the MoS in AP direction is directly dependent on walking speed this systematic speed difference explained the difference in AP MoS between self-paced walking and walking at fixed speed\textsuperscript{59, 98}.

The adaptations in step length, frequency and width in response to the perturbations suggest that these adaptations are part of a strategy to decrease the probability of falling. Nevertheless, these adaptations did not prevent $\lambda_{5\text{-step}}$ to increase, with increasing perturbation intensity. This result implies that the perturbations caused an increased risk of falling, which is in line with the results found by McAndrew et al.\textsuperscript{86}. At the same time we found that subjects increased their backward and sideward MoS, which implies a decrease in risk of falling for these directions. This seems to be contradictory, but subjects probably created a sufficiently wide margin within which a decrease in local dynamic stability can be allowed, without increasing the risk of falling. This is in agreement with the results of the study of Hof et al.\textsuperscript{58}, who found larger ML MoS for above-knee amputees during walking compared to healthy controls, although amputees are locally less stable during walking than able-bodied people\textsuperscript{77}.

The increases in backward and sideward MoS in response to the perturbations are a direct consequence of the adaptations in step length, step frequency and step width. A basic requirement of walking is that the COM passes the stance foot during each single support phase. Otherwise a backward fall will occur. Consequently, the XCoM should always be in front of the dorsal border of the BoS\textsuperscript{38, 98}. Pai and Patton demonstrated that decreasing step length, in combination with an unchanged walking speed, causes an increase of the MoS in backward direction. In ML-direction subjects should prevent that the XCoM exceeds the lateral border of the BoS. Hof et al. demonstrated that increasing step width and step frequency contribute to an increase in ML MoS\textsuperscript{58}. Therefore step parameters like step frequency, step length and step width, instead of walking speed, seem to be important moderators of the probability of falling.

Although the adaptations in walking pattern in response to the perturbations that were found in this study seem to evoke a clear benefit in terms of increased MoS and therefore can reduce fall risk, it should be noted that they might be specific for the continuous perturbation in ML direction used in this study. The generalization of this response to perturbations in other directions or in situations with discrete perturbations should be further explored.
In conclusion, the results of the present study indicate that the strategy of choice to cope with ML continuous balance perturbations is not a reduction of walking speed, but rather a combination of decreased step length and increased step frequency and step width. As a consequence of the simultaneous decrease in step length and increase in step frequency the unchanged walking speed in response to the perturbations can be regarded as a coincidental result of these adaptations. Although the effect on LDS cannot be inferred from our data, the observed changes in step parameters increase MoS, and therefore seem to decrease the probability of falling.

Conflicts of interest statement
Authors state that no conflicts of interest are present in the research.

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CHAPTER 3

Stepping strategies for regulating gait adaptability and stability

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3.1 Abstract

Besides a stable gait pattern, gait in daily life requires the capability to adapt this pattern in response to environmental conditions. The purpose of this study was to elucidate the anticipatory strategies used by able-bodied people to attain an adaptive gait pattern, and how these strategies interact with strategies used to maintain gait stability.

Ten healthy subjects walked in a Computer Assisted Rehabilitation Environment (CAREN). To provoke an adaptive gait pattern, subjects had to hit virtual targets, with markers guided by their knees, while walking on a self-paced treadmill. The effects of walking with and without this task on walking speed, step length, step frequency, step width and the margins of stability (MoS) were assessed. Furthermore, these trials were performed with and without additional continuous ML platform translations.

When an adaptive gait pattern was required, subjects decreased step length ($p<0.01$), tended to increase step width ($p=0.074$), and decreased walking speed while maintaining similar step frequency compared to unconstrained walking. These adaptations resulted in the preservation of equal MoS between trials, despite the disturbing influence of the gait adaptability task. When the gait adaptability task was combined with the balance perturbation subjects further decreased step length, as evidenced by a significant interaction between both manipulations ($p=0.012$).

In conclusion, able-bodied people reduce step length and increase step width during walking conditions requiring a high level of both stability and adaptability. Although an increase in step frequency has previously been found to enhance stability, a faster movement, which would coincide with a higher step frequency, hampers accuracy and may consequently limit gait adaptability.

Key words
Gait adaptability; Gait stability; Walking speed; Step length; Step frequency
3.2 Introduction

For patients with gait impairments, falling during walking is one of the main problems\textsuperscript{92-93}. Knowledge of strategies that patients could use to minimize the risk of falling is essential in order to evaluate and improve this aspect of walking in rehabilitation. Studies in which fall risk is manipulated in able-bodied people may help to discover these strategies. Multiple studies have used balance perturbations to investigate how people respond to situations in which gait stability is decreased\textsuperscript{46, 85-86}. However, being stable is not the only criterion for preventing falls. People also have to be able to adapt this stable gait pattern to comply with discrete changes in environmental conditions (e.g. to avoid an obstacle or to select safe foot placement locations).

In recent studies, continuous medio-lateral (ML) surface translations were imposed in healthy subjects, to examine which strategies they use to maintain gait stability\textsuperscript{46, 85-86}. As changing walking speed has been proposed to be an important strategy to enhance gait stability\textsuperscript{31, 37, 112, 124}, we previously\textsuperscript{46} used a self-paced treadmill allowing subjects to adapt their walking speed in response to the perturbations. Unexpectedly, subjects did not decrease walking speed in response to the perturbations. Instead, they shortened step length and increased step frequency, while keeping walking speed constant. These adaptations can be explained as strategies to increase the margin of stability (MoS) in medio-lateral (ML) and backward (BW) direction, and therefore decrease the risk of falls in these directions\textsuperscript{57-58}.

Since an increase in BW MoS directly implies a decrease in the forward MoS, it appears that decreasing the risk of a backward fall is preferred above decreasing the risk of a forward fall when stability of walking is challenged\textsuperscript{46}.

Studies that used protocols to provoke an adaptive gait pattern mostly used an obstacle avoidance task\textsuperscript{27, 60, 107-108}. In some of these studies\textsuperscript{27, 60} obstacles were presented with limited preview time, and therefore quick discrete adaptations of the gait pattern were necessary to avoid these obstacles. Expectation or fear for these suddenly appearing obstacles, however, might also elicit anticipatory adaptation in the steady state gait pattern to be able to better execute a fast and accurate movement in response when necessary, while simultaneously ensure dynamic stability in response to the destabilizing effect of this movement\textsuperscript{21, 107-108}.

The purpose of this study was to investigate which strategies, in terms of spatio-temporal gait parameters, are used by able-bodied people to enhance gait adaptability in a situation that requires an accurate and fast adaptation with a limited response time. This was done by using a gait adaptability task (GA-task), in which subjects had to hit virtual targets that were projected on a 2D screen, using virtual markers controlled by their knee motion. We hypothesized that subjects need to increase or at least maintain their BW and ML MoS to
withstand the disturbing influence of these fast adaptations. Therefore, based on previous experiments, we hypothesized that subject would decrease step length and increase step width in response to the GA-task\textsuperscript{46, 86}. An increase in step frequency, previously found as strategy to enhance stability\textsuperscript{46, 86}, will imply less time to plan and execute the movement of the knee, necessary to hit the target, and might hamper the accuracy of this movement\textsuperscript{10, 20, 41}. Therefore, we hypothesized that an adaptation in step frequency might not be selected during the GA-task. To further study this potential conflict between gait adaptations used to enhance gait adaptability and stability we imposed the GA-task both in the absence and presence of a continuous ML balance perturbation\textsuperscript{46}. We hypothesized that the decrease in step length and the increase in step width in response to the GA-task are larger when the GA-task and the balance perturbations are combined, because of the additive destabilizing effect of the perturbation, while the expected lack of an increase in step frequency during the GA-task would hamper balance control.

3.3 Methods

3.3.1 Subjects
Ten healthy young subjects (8 men and 2 women, age: 23.2±2.9 years, length: 1.82±0.09 m, and weight 71.3±7.5 kg) were included. Subjects provided their written informed consent and the local ethical committee approved the protocol before the experiment was performed.

3.3.2 Equipment
All subjects walked in the Computer Assisted Rehabilitation Environment (CAREN, Motek Medical b.v., Amsterdam, The Netherlands). The CAREN system consists of a treadmill mounted onto a 6-degree-of-freedom motion platform in combination with a Virtual Environment (VE) (figure 3.1a). VE’s were projected onto a 180 degrees cylindrical screen in front of the treadmill. During all trials, a virtual road surrounded by trees was projected to create an optical flow pattern. The VE was also used to project targets that had to be hit during the GA-task (figure 3.1b). The motion platform was used to induce perturbations in ML-direction in selected trials (figure 3.1c). D-flow software was used to control the system and to synchronize the instantaneous treadmill speed and scene progression\textsuperscript{43}. Twelve high resolution infra-red cameras (Vicon, Oxford, UK) and the Vicon Lower Body Plug-in-Gait marker set were used to capture kinematic data. The treadmill was used in the self-paced mode, which allowed subjects to modify walking speed at will and adapt speed to the manipulations imposed during this experiment. This was done by servo-controlling the motor with a real-time algorithm that took into account the anterio-posterior (AP) position.
of the pelvis markers and the AP-midline of the treadmill. A safety harness system suspended overhead prevented the subjects from falling, but did not provide weight support.

**Figure 3.1**: (A) Experimental setup: CAREN and Virtual scene; (B) GA-task with an example of a target in the right panel. The white dots represent a projection of the knee markers. The distance between the knee marker, in this example the left knee marker, and the centre of the target was used as outcome measure for the accuracy of the knee movements while performing the gait adaptability task; (C) ML balance perturbation with the perturbation pattern in the right panel.
### 3.3.3 Protocol

**Familiarisation**

Subjects performed five familiarisation trials of 2 min each, one trial at fixed walking speed and one trial for each experimental condition. Purpose of these trials was to become familiar with walking on a (self-paced) treadmill, the VE and the various manipulations.

**Experimental trials**

The protocol consisted of four trials of 4 minutes walking: (1) a trial of unperturbed walking, (2) a trial with a GA-task, (3) a trial with continuous balance perturbations, and (4) a trial with both a GA-task and balance perturbations. The first minute of each trial was used to let subjects get familiar to the self-paced setting of the treadmill and the different manipulation. All trials were offered in random order.

For the GA-task, the VE was used to project targets on the screen. In addition, a projection of the markers attached to the knees was shown on the screen. Subjects were instructed to hit the targets on the screen with the projected knee markers, as close as possible to the centre of the targets. In each trial, a total of 32 targets appeared. Targets appeared at initial contact and disappeared after the duration of one gait-cycle, which was estimated from the gait pattern during the first minute of the trial. The positions of the targets differed randomly in side (left or right), height (120% or 140% of lower leg length), and ML-position (120% or 140% of distance between the left and right anterior superior iliac spines from the midline of the treadmill) to increase the level of unpredictability of this task.

For the balance perturbations, translations of the walking surface in ML-direction were used, following a multi-sine function:

\[ D(t) = 0.05[1.0 \sin(0.16 \cdot 2\pi t) + 0.8 \sin(0.21 \cdot 2\pi t) + 1.4 \sin(0.24 \cdot 2\pi t) + 0.5 \sin(0.49 \cdot 2\pi t)] \]

where \( D(t) \) is the translation distance (m) and \( t \) is time (s) \(^{46,85-86} \).

### 3.3.4 Data collection

To determine walking speed, the speed of the treadmill was recorded. Kinematic data of markers attached to the lateral malleoli of the ankles and the lateral epicondyles of the knees were collected with a Vicon system. Both speed and kinematic data were recorded with the D-flow software. The sample rate was 100 samples/s and before data analysis kinematic data were bi-directionally low-pass filtered (Butterworth, 10 Hz cut-off). The final 3 min of each trial were used for data analysis.
3.3.5 Data analysis

Walking speed
Walking speed was calculated as the average treadmill speed over the final 3 min of each trial.

Step parameters
Step frequency was determined as the inverse of the average duration between two subsequent heel-strikes, where heel-strikes were detected as the local maxima of the AP position of the markers attached to the lateral malleoli. Step width was calculated as ML-distance between both ankle markers at the instant of heel-contact and step length was defined as the AP-distance between these markers at the instant of heel-contact.

Margins of stability
To calculate the margins of stability (MoS), a method derived from the procedure developed by Hof was used. In this study, the extrapolated centre of mass (XCoM), was defined as the average of four markers attached to left and right anterior and posterior superior iliac spines, as estimate for the CoM, plus its velocity times a factor $\sqrt{l/g}$, with $l$ being the maximal height of the estimated CoM and $g$ the acceleration of gravity. The MoS were calculated for both the BW and ML-direction as the distance between the lateral malleolus of the ankle of the leading foot (representing the border of the base of support) and the position of the XCoM for the moment at which the MoS reached its minimum value within each step. Although basically similar our method differs from Hof who used force plate data for calculating XCoM and MoS. Besides, in our definition of the BW MoS, the equation XCoM-BoS was used, instead of the original equation BoS-XCoM by Hof et al.

Gait adaptability
Gait adaptability was quantified by the performance on the GA-task. This performance is defined as the minimum Euclidean distance between knee projection and target centre.
Figure 3.2: Schematic representation of the calculation of the ML (A) and BW (B) MoS. Trajectories of the margin of the BoS (solid line), CoM (dashed line), and XCoM (dotted line) are shown for a period of approximately two steps. The MoS is the difference between the trajectory of the XCoM and the margin of the BoS for the moment at which the MoS reached its minimum value within the period of one step (represented by the arrows).

3.3.6 Statistical design
To measure the effects of the GA-task and the balance perturbation on step length, step frequency, step width, walking speed, and the MoS in ML and BW direction, 2 × 2 within-within factorial ANOVAs were performed, with the absence or presence of the GA-task and the absence or presence of the balance perturbation as within factors. P-values less than 0.05 were considered significant. When an interaction between the effect of the GA-task and the effect of the perturbation was present, paired-samples t-tests with a Bonferroni correction (critical p-value: 0.025) were performed to investigate for which level of perturbation the effect of the GA-task was significant. A paired-samples t-test was used to compare the performance between the GA-task with and without perturbation. Statistical analyses were performed using SPSS 17.0 (SPSS Inc., Chicago, IL).
3.4 Results

The results of the statistical analyses are reported in table 3.1. Step length decreased significantly in the presence of both the GA-task and the balance perturbation (Figure 3.3a). In addition, a significant interaction effect of the GA-task and the balance perturbation on step length was found. The results of the post-hoc test showed that step length was significantly affected by the GA-task for both conditions without \((t=3.780; \ p=0.004; \ df=9)\) and with \((t=6.974; \ p<0.001; \ df=9)\) perturbation, but this effect was larger with perturbation. For step width a main effect of balance perturbation was found. In response to the GA-task, step width was slightly increased. Besides, this increase appeared larger for the condition with perturbation, but both effects were not significant (Figure 3.3b). Step frequency was not affected significantly by either the GA-task or the balance perturbation and no significant interaction was found (Figure 3.3c). A main effect on walking speed was found for both the GA-task and the balance perturbation. With both manipulations walking speed decreased. There was a tendency towards an interaction effect between the GA-task and the balance perturbation, but this was not significant (Figure 3.3d).

For one of the subjects, quality of the kinematic data of the pelvis was not sufficient to estimate the trajectory of the centre of mass. Therefore MoS were calculated for nine subjects. ML MoS did not change significantly in response to the balance perturbation or the GA-task (Figure 3.4a). The BW MoS increased in response to the balance perturbation, but was not affected significantly by the GA-task. No significant interaction effects were found for both ML and BW MoS (Figure 3.4b). The average distance between knee projection and target, as outcome measure of the performance on the GA-task, was 3.29 (±0.46) cm for the condition without balance perturbation and 3.47 (±0.48) cm for the condition with balance perturbation, which was not significantly different between conditions \((t=0.371; \ p=0.720; \ df=9)\).
Table 3.1: Results for the statistical analyses

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Main effect GA-task</th>
<th>Main effect perturbation</th>
<th>Interaction GA-task × perturbation</th>
</tr>
</thead>
<tbody>
<tr>
<td>Step length</td>
<td>F = 65.182 p &lt; 0.001* df=1,9</td>
<td>F = 46.517 p &lt; 0.001* df=1,9</td>
<td>F = 9.941 p = 0.012* df=1,9</td>
</tr>
<tr>
<td>Step width</td>
<td>F = 4.095 p=0.074 df=1,9</td>
<td>F = 6.001 p=0.037* df=1,9</td>
<td>F=2.407 p=0.155 df=1,9</td>
</tr>
<tr>
<td>Step frequency</td>
<td>F=0.604 p=0.457 df=1,9</td>
<td>F=2.757 p=0.131 df=1,9</td>
<td>F=0.548 p=0.478 df=1,9</td>
</tr>
<tr>
<td>Walking speed</td>
<td>F=7.622 p=0.022* df=1,9</td>
<td>F=15.441 p=0.003* df=1,9</td>
<td>F=3.102 p=0.112 df=1,9</td>
</tr>
<tr>
<td>ML MoS</td>
<td>F=0.885 p=0.374 df=1,8</td>
<td>F=0.355 p=0.568 df=1,8</td>
<td>F=3.823 p=0.086 df=1,8</td>
</tr>
<tr>
<td>AP MoS</td>
<td>F=1.920 p=0.203 df=1,8</td>
<td>F=5.525 p=0.047* df=1,8</td>
<td>F=0.023 p=0.883 df=1,8</td>
</tr>
</tbody>
</table>

* Significant at the 0.05 level
Figure 3.3: Average and standard deviation of step length (A), step width (B), step frequency (C), and walking speed (D) \((n = 10)\) for all experimental conditions. The black symbols represent the trials with perturbation and the white symbols represent the trials without perturbation. GA stands for the gait adaptability task. Significant main effects of the GA-task and the perturbation are indicated with respectively * and +. Significant GA-task × perturbation effects are indicated by ‡.
3.5 Discussion

The purpose of this study was to elucidate which strategies, in terms of spatio-temporal gait parameters, are employed by able-bodied people to attain an adaptive gait pattern, and how these strategies interact with strategies used to maintain gait stability. To provoke an adaptive gait pattern a GA-task was used, which required quick and accurate adaptations of the gait pattern, with little preparation time. In line with our hypothesis, step frequency was not adapted during the GA-task, whether perturbations were present or not. Step length decreased in response to the GA-task, and this decrease was significantly larger for the condition in which the GA-task was combined with the balance perturbation, which was again in agreement with our hypothesis. Besides, this decrease in step length also resulted in a decrease in walking speed during the trials with GA-task, compared to normal walking. With respect to step width, only a tendency towards an increase in step width in response to the GA-task was found, which does not fully support our hypothesis that step width would be increased to preserve the ML MoS.

The imposed GA-task required discrete fast and accurate adaptations of the steady state gait pattern. At the same time, such an adaptation might induce a loss of balance, since the required knee movement affects the body centre of mass movement trajectory\textsuperscript{21, 107-109}. The results of this study showed that subjects were able to maintain their BW and ML MoS, despite the disturbing influence of the GA-task. The preservation of the MoS is likely related to the adaptations observed in the steady state gait pattern, i.e. decreasing step length and the tendency towards an increase in step width, which have previously been identified as
mechanisms to respectively increase the BW and ML MoS to 46, 57-58, 85, 98. As can be appreciated from Figure 3.5, step length and step width during the trials with the GA-task did not just differ from normal walking for the steps in which the target had to be hit, but especially also for the steps in between the targets. We therefore conclude that to preserve gait stability during the adaptability task subjects anticipated to the appearance of the targets by decreasing their step length and increasing step width of their steady state gait pattern. This allowed making a quick adaptation of the gait pattern, when the target actually appeared, without losing balance.

In our previous study, we also found an increase in step frequency in the presence of a balance perturbation 46. In the current study however, this strategy was not observed during the GA-task, either with or without balance perturbation. Apparently, the gait pattern required for performing the GA-task imposed conflicting demands on step frequency. The GA-task required the subjects to make an accurate goal directed response to an environmental cue. According to Fitts’ law 41, the accuracy of a movement is positively correlated with the available movement time. An increase in step frequency during the GA-task would imply less time to plan and execute the movement of the knee 10, 20, which likely has a negative influence on the performance of this task 36, 41. The lack of an increase in step frequency during the GA-task indicates that, the young and healthy subjects who participated in the current study, apparently prioritised performance on the GA-task above the preservation of stability. The subjects seem to compensate for this detrimental effect of the absence of a decrease in step frequency on gait stability by further decreasing step length, especially during the trials in which the disturbing effect of the GA-task was strengthened by the balance perturbation. We have hypothesised a similar effect on step width, however, only a small tendency towards an interaction between the GA-task and the balance perturbations on step width was found.

Although the adaptations found in the current study seem to be small, the magnitude of the adaptations in step length and walking speed in response to the GA-task lies within the range of differences found between, for example able-bodied people and people walking with a lower limb prosthesis 23, 99, a group at an increased risk of falling 92-93. The adaptations of these parameters were even larger when the GA-task was combined with the perturbation. The magnitude of the responses in this study can therefore be regarded as functionally relevant.

In the present study, only a slight tendency towards an increased step frequency in response to the balance perturbation was observed. Because step frequency has a positive effect on the ML MoS, this could also be the reason why the ML MoS did not increase in response to the perturbation 50. This appears to be in contrast with the results of our previous study,
where we found an increase in step frequency and ML MoS in response to the same balance perturbation. This disparity might be caused by differences in the protocols of both studies. In our previous study, subjects also underwent trials with perturbations of much larger amplitudes than in the present study. The larger perturbations may have had a cross-over effect on the strategies chosen by the subjects during the perturbations with lower amplitudes. Other explanations could be a lower statistical power, or the lower age of the subjects in the current study. Although the subjects in our previous study could not be considered as 'older adults', they were, on average, 9 years older.

A limitation of this study is the estimation of the CoM as the average of the pelvis markers to calculate the XCoM. This is not an exact representation of the CoM, but errors made through this approximation were likely similar for all conditions, and might therefore not affect differences in MoS between conditions.

In conclusion, able-bodied people reduced step length, tended to increase step width, but kept step frequency constant during walking conditions that required a high level of both stability and adaptability. The decrease in step length and the trend towards an increase in step width probably ensured sufficiently large BW and ML MoS when performing the GA-task. Although an increase in step frequency has previously been shown to be selected when stability is perturbed, it will limit the available response time and hence accuracy of adapted movements. This may explain why step frequency was not adjusted while performing the GA-task. As a result of the decrease in step length and an unchanged step frequency, walking

**Figure 3.5:** A typical example for the pattern of step length (upper panel) and step width (lower panel) for 100 steps of normal walking (grey bars), and 100 steps of a trial with GA-task (white bars). In this example 98 steps are smaller and 75 steps are wider during walking with the GA-task than during normal walking.
speed decreased in such situations. It should be noted that the strategies found in the present study are specific for the situation in which a quick and accurate adaptation of the gait pattern is required. Generalisation of these results to situations with different temporal and spatial constraints should be treated with caution. Nevertheless, these data show how people cope with environmental constraints in the real world by adapting spatio-temporal step parameters. The next step is to investigate whether people with gait impairments who are at risk of falling have the ability to use these strategies to a same degree as the healthy population participating in this study.

*Conflicts of interest statement*
Authors state that no conflicts of interest are present in the research.

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CHAPTER 4

Steps to take to enhance gait stability: The effect of stride frequency, stride length, and walking speed on local dynamic stability and margins of stability

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4.1 Abstract

The purpose of the current study was to investigate whether adaptations of stride length, stride frequency, and walking speed, independently influence local dynamic stability and the size of the medio-lateral and backward margins of stability during walking. Nine healthy subjects walked 25 trials on a treadmill at different combinations of stride frequency, stride length, and consequently at different walking speeds. Visual feedback about the required and the actual combination of stride frequency and stride length was given during the trials. Generalized Estimating Equations were used to investigate the independent contribution of stride length, stride frequency, and walking speed on the measures of gait stability.

Increasing stride frequency was found to enhance medio-lateral margins of stability. Backward margins of stability became larger as stride length decreased or walking speed increased. For local dynamic stability no significant effects of stride frequency, stride length or walking speed were found.

We conclude that adaptations in stride frequency, stride length and/or walking speed can result in an increase of the medio-lateral and backward margins of stability, while these adaptations do not seem to affect local dynamic stability. Gait training focusing on the observed stepping strategies to enhance margins of stability might be a useful contribution to programs aimed at fall prevention.
4.2 Introduction

While substantial variance in walking speed is apparent between individuals, a more or less consistent speed is selected by individuals during steady state walking\textsuperscript{115}. Moreover, at any given walking speed it is possible to select different combinations of stride frequency and stride length, but again individuals tend to choose a specific stride frequency and length consistently\textsuperscript{76}. Previous studies have suggested that energy cost\textsuperscript{5-6} and variance thereof due to variance in anthropometrical characteristics such as leg inertia\textsuperscript{76, 115} determine the selection of a certain gait pattern. However, optimizing energy cost might not be the sole objective when walking. Especially for people at an increased risk of falling, the choice for a certain gait pattern may be related to improving gait stability and reducing fall risk rather than minimizing energy cost. Fallers and fall-prone people often walk slower, with shorter steps and a lower step frequency than non-fallers\textsuperscript{23, 99, 122, 124}. These differences in gait pattern, in particular the lower walking speed, are often explained as strategies to decrease fall risk\textsuperscript{31, 37, 70, 75}. However, from such observational data, it remains unclear whether these differences in gait pattern represent a strategy to reduce the risk of falling or whether these differences in gait pattern serve other purposes, and might even coincide with an increased risk of falling.

To investigate whether gait pattern selection or adaptations could serve the purpose of decreasing fall risk, several studies have examined responses to balance perturbations during gait\textsuperscript{46-48, 85, 87, 90}. Evidence was found that able-bodied people, but also people with a trans-tibial prosthesis\textsuperscript{48}, effectively deal with perturbations by increasing their stride frequency and decreasing their stride length, while keeping walking speed constant. These results suggest that instead of a decrease in walking speed, an increase in stride frequency and a decrease in stride length are adopted to minimize the risk of falling.

An alternative approach to investigate whether selection of spatio-temporal step parameters could serve the purpose of reducing fall risk is to systematically examine their effects on gait stability. In contrast to the first approach, this approach is dependent on the selected parameter to quantify gait stability or associated fall risk. The short-term Lyapunov exponent ($\lambda_s$), a measure of the local dynamic stability (LDS), was proposed to quantify gait stability\textsuperscript{28} and has gained considerable support\textsuperscript{17}. The $\lambda_s$ quantifies the average logarithmic rate of divergence of a system after a small perturbation over a period of 0.5 or 1 stride. An increase in $\lambda_s$ thus implies a decrease of LDS. Previous studies indicate that the margins of stability (MoS) may provide additional information on gait stability\textsuperscript{46, 57, 98}. The size of the MoS reflects the divergence of the centre of mass (CoM) that one can handle before an actual loss of balance occurs. The MoS is calculated as the distance between the extrapolated centre
of mass (XCoM) and the limits of the base of support, in which the XCoM represents the state of the CoM taking into account both its position and velocity\textsuperscript{57}. MoS can be calculated in medio-lateral (ML; figure 1.4a) and backward (BW; figure 1.4b) direction. A negative ML MoS will result in a deviation from the straight walking trajectory, while a negative BW MoS will result in an interruption of the forward progression. Consequently, in case of a negative MoS, a crossing step or a backward step will be necessary to prevent, respectively, a sideward or backward fall. It should be noted that an increase in BW MoS simultaneously has the disadvantage that the risk of a forward loss of balance will increase. However, while a backward loss of balance inflicts a large chance of state of the walking movement, requiring a reversal of the periodic leg movement (stepping backward) to regain balance, a forward loss of balance requires a relatively small adaptation of the next step(s), like a temporary increase in step length, to recover\textsuperscript{59}. In line with the latter, we have previously found that able-bodied subjects, but also transtibial amputees preferred to increase their BW MoS in response to ML perturbations of the walking surface\textsuperscript{46, 48}.

Several studies have investigated the effect of walking speed and the underlying step parameters on $\dot{\lambda}s$. Although most of these studies found that a decrease in walking speed decreased the $\dot{\lambda}s$, and therefore increased LDS\textsuperscript{29, 31, 37, 70}, a more recent study, in which the effect of time-series length and number of steps on $\dot{\lambda}s$ calculation was taken into account, disputed this claim\textsuperscript{13}. McAndrew and Dingwell\textsuperscript{89} investigated the effect of different stride lengths at the same walking speed on $\dot{\lambda}s$ of trunk kinematics. They expected to find lower values for $\dot{\lambda}s$ when subjects walk with short steps. However, this appeared to be true for only the $\dot{\lambda}s$ calculated for the medio-lateral direction, and therefore these results did not fully reconcile earlier conflicting findings on the relationship between adaptations in step length and fall risk.

In contrast with the effects of the gait pattern on LDS, the effect of adaptations of the gait parameters on the size of the ML and BW MoS can be predicted analytically. Based on the inverted pendulum behavior of the human body while walking, an increase in stride frequency is expected to increase the ML MoS\textsuperscript{57, 59}, while a decrease in stride length and an increase in walking speed is expected to increase the BW MoS\textsuperscript{38-39}. In agreement with these models an increase in BW MoS was indeed found when subjects walked faster or with shorter steps compared to their comfortable gait pattern\textsuperscript{38-39, 88}.

The interpretation of the experimental results described in the previous paragraphs is, however, obscured by the fact that stride frequency, stride length, and walking speed cannot be adapted independently from each other. Consequently, the results might not only be caused by the imposed manipulation, but could also be an effect of a concomitant change in
one of the other gait parameters. Therefore, a systematic analysis of the effect of stride length and frequency and their interaction, i.e., walking speed, should be conducted to reveal their independent effects on gait stability and associated fall risk.

The purpose of the current study was to investigate whether manipulations of stride frequency, stride length, and consequently walking speed, independently influence LDS expressed as \( \lambda_s \) and the ML and BW MoS. We hypothesized that \( \lambda_s \) would change in response to the imposed manipulations, but because of the large variation in results of previous studies that investigated the effect of gait pattern manipulations on \( \lambda_s \), the direction of the change could not be predicted. With respect to the MoS, we hypothesized that the ML MoS only increases due to an increase in stride frequency\(^{56} \), while BW MoS increases due to a decrease in stride length and due to an increase in walking speed\(^{38-39} \).

### 4.3 Methods

#### 4.3.1 Subjects

Nine young healthy subjects (6 men and 4 women, age: 21.9 ± 1.8 years, weight: 73.4 ± 8.7 kg, length: 1.79 ± 0.09 m) were included. The study was approved by the local ethical committee of the Faculty of Human Movement Sciences, VU University Amsterdam, and subjects gave written informed consent prior to their participation.

#### 4.3.2 Equipment

All subjects walked on an instrumented treadmill placed in the front of a screen, which was used to give subjects real-time, visual feedback about the combination of stride frequency and stride length they were employing (Figure 4.1). Single infrared Light Emitting Diodes (LEDs) were attached to the lateral malleoli of the ankles and the heels of each subject, and a neoprene band with a cluster of three LEDs was attached between the left and right posterior superior iliac spines (PSIS). The LED’s were used for movement registration with an active 3D movement registration system (Optotrak® Northern Digital Inc., Waterloo, Ontario, Canada). A custom made application (LabVIEW, National Instruments, Utrecht, The Netherlands) was used to calculate stride frequency and stride length during walking, based on the position of the markers attached to the ankles, and to give real time feedback about these parameters.
4.3.3 Protocol

The experiment started with a pre-experimental trial. In this trial the comfortable walking speed was determined, by gradually increasing the belt speed until a comfortable walking speed was reported. Belt speed was then increased beyond the reported comfortable walking speed and subsequently gradually decreased again until a comfortable walking speed was reported. Comfortable walking speed was determined as the average of the two reported comfortable walking speeds. After the determination of comfortable walking speed, the treadmill was set at this walking speed to determine comfortable stride frequency and stride length over a period of 2 minutes.

For the experimental protocol subjects were asked to walk at five different stride frequencies (columns in figure 4.2) in combination with five different stride lengths (rows in figure 4.2), expressed as a percentage of the comfortable values. This resulted in 25 trials of 4 minutes walking, all at an unique combination of stride frequency and stride length. The percentages...
of comfortable stride frequency and stride length were chosen in such a way that walking speed was the same for the trials on the diagonals from the upper left to the lower right in the experimental overview presented in figure 4.2. For each trial, treadmill speed was set at the required speed. Visual feedback on the required and the current combination of stride frequency and stride length was given as shown in figure 4.1. Subjects were instructed to keep the red dot as much as possible in the middle of the light gray cell. All trials were divided over two days and were offered in random order.

Figure 4.2: Schematic overview of the experimental conditions. (V: Walking speed; SL: Stride length, SF: Stride frequency). Percentages are percentages of respectively comfortable walking speed, stride length, and stride frequency. The rows represent the stride length manipulations and the columns the stride frequency manipulation. Note that walking speed on the diagonals reached a constant percentage of comfortable walking speed.
4.3.4 Data collection

Kinematic data of markers attached at the lateral malleoli of the ankles, the heels and the cluster markers placed between the left and right PSIS were collected with the Optotrak system at a sampling rate of 100 samples/s.

4.3.5 Data analysis

Before data analysis, except for the calculation of LDS, kinematic data were low-pass filtered with a bi-directional Butterworth filter with a cut-off frequency of 4 Hz. All parameters were calculated for the first 150 strides after the first 15 seconds of each trial.

Stride frequency and stride length

Stride frequency was determined as the inverse of the average duration between two subsequent heel-strikes, where heel-strikes were detected as the local maxima of the position of the ankle markers in the anterio-posterior direction. Stride length was calculated as the anterio-posterior distance between both ankle markers at the moment of heel-contact.

Margins of stability

To calculate the margins of stability, a method derived from the procedure developed by Hof et al. was used. In the present study, the extrapolated centre of mass (XCoM), was defined as the average of the three cluster markers attached between the left and right PSIS, as estimate for the centre of mass (CoM), plus its velocity times a factor \( \sqrt{l/g} \), with \( l \) being the maximal height of the estimated CoM and \( g \) the acceleration of gravity. The margins of stability (MoS) were calculated as the position of the XCoM relative to the lateral malleolus of the ankle of the leading foot for the sideward MoS and relative to the heel marker of the leading foot for the backward MoS. For each step sideward and backward MoS were calculated for the moment at which the MoS reached its minimum value within each step and were subsequently averaged over 150 strides (300 steps). Although basically similar, our method differs from that of Hof et al. who used force plate data for calculating XCoM and margins of stability. Besides, in our definition of the BW MoS, the equation XCoM-BoS was used, instead of the original equation BoS-XCoM by Hof et al.

Local dynamic stability

To calculate LDS, position data of the cluster marker attached to the pelvis was used. Data were analyzed without filtering. Linear velocities in the ML, AP, and vertical (VT) direction were calculated as the first derivative of the position of the average of the cluster marker. Time-series were time-normalized, using a shape-preserving spline interpolation, such that each time-series of 150 strides had a total length of 15,000 samples. Subsequently, 9D state spaces were reconstructed from the time-normalized 3D linear velocity time series, each with their 25 and 50 samples time delayed copies.
From the constructed state spaces, Euclidean distances between neighbouring trajectories in state space were calculated as a function of time and averaged over all original nearest neighbour pairs to obtain the average logarithmic rate of divergence. The slope of the resulting divergence curves for the interval between 0-50 samples provides an estimate of $\lambda s$ for the period of approximately 0-1 step$^{11, 104}$. Larger values of $\lambda s$ imply that the LDS of the gait pattern is lower.

4.3.6 Statistical analysis

To establish the relationship between stride frequency, stride length, and walking speed, all expressed as a fraction of the comfortable value, on the one hand, and the gait stability measures (ML and BW MoS, and the $\lambda s$) on the other hand, Generalized Estimating Equations (GEE) were used. GEE is a regression analysis technique that accounts for the dependency of the repeated measurements. Because all trials were offered randomly, an exchangeable working correlation matrix was chosen to define this dependency of the repeated measurements in the model. With this technique it is possible to determine independently the contribution of stride frequency, stride length, and walking speed to the different outcome measures. This analysis was performed using IBM SPSS Statistics 20.0.

4.4 Results

Figure 4.3 graphically depicts the results for ML and BW MoS, as well as $\lambda s$, for all 25 experimental trials. The results of the GEE analyses are summarized in table 4.1. For ML MoS only a positive effect of an increase in stride frequency was found. BW MoS became larger as a result of a decrease in stride length and as a result of an increase in walking speed. For $\lambda s$ no significant effects of stride frequency, stride length or walking speed were found.
Figure 4.3: Results for ML MoS, BW MoS, and LDS. ML MoS (upper row), BW MoS (middle row), and λ.s (lower row) as a function of both stride length and stride frequency (left column). Light (yellow) areas represent a high value, while a dark (red) areas represent a low value for the concerning outcome measure. In the middle column the figure is oriented such that the relation between stride frequency and the outcome measure stands out. The same is done for stride
Table 4.1: GEE regression coefficients (β) for the effect of stride length, stride frequency and walking speed on the one hand and the ML MoS, the BW MoS and LDS expressed as λs on the other hand. The model used is: outcome measure = β₁*stride length + β₂*stride frequency + β₃*walking speed + intercept. In the model stride length, stride frequency, and walking speed are expressed as the fraction of the comfortable value.

ML MoS:

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<tr>
<th>Parameter</th>
<th>β</th>
<th>p-value</th>
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<tbody>
<tr>
<td>Stride length</td>
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<td>Stride frequency</td>
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<tr>
<td>Walking speed</td>
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<td>0.438</td>
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BW MoS:

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<th>p-value</th>
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<td>0.009*</td>
</tr>
<tr>
<td>Stride frequency</td>
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<td>0.716</td>
</tr>
<tr>
<td>Walking speed</td>
<td>0.465</td>
<td>&lt;0.001*</td>
</tr>
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</table>

LDS (λs):

<table>
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<th>p-value</th>
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<tbody>
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<td>Stride length</td>
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<td>0.084</td>
</tr>
<tr>
<td>Stride frequency</td>
<td>0.288</td>
<td>0.331</td>
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<tr>
<td>Walking speed</td>
<td>-0.546</td>
<td>0.060</td>
</tr>
</tbody>
</table>

* significant at the 0.05 level

4.5 Discussion

The purpose of the present study was to investigate whether adaptations in stride frequency, stride length, and walking speed independently influence λs and ML and BW MoS, during walking. The results obtained contribute to the understanding of gait pattern selection in healthy and patient populations. Moreover they provide pointers for gait strategies that might be used in training programs aimed at fall prevention. We found that ML MoS increased with an increase in stride frequency, while adaptations of stride length and walking speed did not affect ML MoS. BW MoS increased with a decrease in stride length or an increase in walking speed, while stride frequency did not affect the BW MoS. Finally, none of the manipulations of stride frequency, stride length, and walking speed significantly influenced λs.
The increase in ML MoS with an increase in stride frequency and the increase in BW MoS with a decrease in stride length or an increase in walking speed were in line with our hypotheses. These results support the applicability of the theoretical models with respect to the MoS, in which the human body during walking is modeled as a simple inverted pendulum. Furthermore, the results support previous observations that the increase in stride frequency and decrease in stride length found in several perturbation studies are strategies to increase ML and BW MoS, and thus to prevent a fall in the presence of perturbations.

The positive effects of a high walking speed and a small stride length on BW MoS may result in a conflict. Based on the representation of the results in figure 4.3 (right panel), we can conclude that the positive effect of the increase in walking speed on BW MoS outweighs the negative effect of the increase in stride length. This can also be derived from the larger regression coefficient ($\beta$) for walking speed, compared to stride length (table 1). For example, an increase in walking speed from 100% to 120% of comfortable walking speed will increase the BW MoS by 0.093 m. (1.2 × 0.465 compared to 1.0 × 0.465), while an increase from 100% to 120% of comfortable stride length will only decrease the BW MoS by 0.009 m. (1.2 × -0.045 compared to 1.0 × -0.045). So increasing walking speed enhances MoS even if it is achieved through an increase of stride length.

Remarkably, none of the manipulations of stride frequency, stride length, and walking speed had a significant effect on $\lambda_s$, although tendencies towards a significant increase in $\lambda_s$ with an increase in step length, and a decrease in $\lambda_s$ with an increase in walking speed were found. Besides, the regression coefficients ($\beta$) suggest that the effects of stride frequency and stride length are opposite to the effect of walking speed, which means that these effects might cancel each other when increasing or decreasing walking speed (table 4.1). Previous studies have shown that LDS, quantified as $\lambda_s$, did decrease as a result of external perturbations, galvanic vestibular stimulation, or as a consequence of gait-impairments, like a lower-limb amputation. However, the present results indicate that LDS does not change when people deviate from their comfortable gait pattern. The $\lambda_s$ reflects the response to (small) perturbations within a period of one step after such a perturbation. This time period might be too short to allow influence on $\lambda_s$ by adaptations of stride frequency and stride length which logically sort an effect after a period of a full stride. Consequently, $\lambda_s$ might be more dependent on intrinsic stiffness and gains of rapid feedback loops.

In comparing the present results on $\lambda_s$ to the results of previous studies, the methodological choices made with respect to the calculation of $\lambda_s$ have to be taken into account. The choices with respect to filtering and state space reconstruction might have influenced the results as
different effects of gait speed and stride length on $\lambda_s$ were previously found for different state space representations of trunk kinematics (i.e.\textsuperscript{12, 89}). However, we have tested the effect of filtering of the data before calculating $\lambda_s$ (10 Hz low-pass filter), and the effect of calculating $\lambda_s$ separately for different planes (ML, AP and VT-plane, in line with McAndrew et al.\textsuperscript{89}), and observed these alternative methods would not have influenced the conclusions drawn in the current study. These findings are in line with a previous study in which it appeared that the correlation between $\lambda_s$ calculated from various state spaces was high\textsuperscript{116}. Nevertheless, it should be taken into account that generalization of the conclusions drawn in the current study with respect to $\lambda_s$ should be done keeping in mind these methodological issues.

For a proper interpretation of the results of this study, it is also important to note that our definition of MoS in anterior-posterior direction differ the from definition previously used by Hof\textsuperscript{59}. Hof defined stability in anterio-posterior direction as the ability to maintain a more or less steady forward speed, while our definition for stability in anterio-posterior direction is the maintenance of forward progression. In the latter case the minimum criterion to maintain stability in anterio-posterior direction is that the XCoM passes the BoS during the double-support phase of the gait cycle, which might not be sufficient to minimize speed decrements and does not exclude speed increases. Secondly, one should be aware that the importance of increasing the BW MoS may strongly depend on the walking condition. In the case of, for example, stepping off a curb\textsuperscript{3}, or walking in a condition with a high risk of a forward trip\textsuperscript{114}, probably decreasing the risk of a forward fall, by increasing the forward MoS will be prioritized above increasing the BW MoS.

In conclusion, in line with the underlying mechanical models, ML MoS can be increased by an increase in stride frequency, and BW MoS can be increased by increasing walking speed or decreasing stride length. It appeared that when walking at a high speed, the positive effect of a high speed on BW MoS outweighs the negative effect of the large stride length employed to reach this high walking speed. When walking speed is limited, for example by energetic constraints, walking with fast and short steps at a certain walking speed results in the largest ML and BW MoS. LDS was not significantly affected by adaptations in stride frequency, stride length, and walking speed. Based on these results, we conclude that adaptations in stride frequency, stride length, and walking speed can be used to increase ML and BW MoS, without a loss of LDS (expressed as $\lambda_s$), when gait stability is challenged. Increases in ML and BW MoS may thus compensate for a potential decrease in LDS, caused by external perturbations\textsuperscript{46, 86} or as a consequence of a gait-impairment\textsuperscript{77}, in order to prevent an actual loss of balance. Besides, the results of the current study are an indication that the slower
walking speed, with a lower stride frequency and a smaller stride length, often employed by people with gait impairments\textsuperscript{23, 99, 122, 124}, may be a cause of an increased fall risk, instead of a strategy used to minimize fall risk. Therefore, it would be of interest to investigate whether people with gait impairments are able to walk at different combinations of stride frequency and stride length, and how these alterations affect the MoS. Training focused on the adaptation in stride frequency and stride length at different walking speeds might help these people to better regulate gait stability, and therewith decrease their risk of falling.

\textit{Conflicts of interest statement}

Authors state that no conflicts of interest are present in the research.

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Walking in an unstable environment: Strategies used by transtibial amputees to prevent falling during gait

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5.1 Abstract

Objective: To investigate which strategies transtibial amputees use to cope with challenges of gait stability and gait adaptability, and how these strategies differ from strategies used by able-bodied controls.

Design: Cross-sectional study

Setting: Computer Assisted Rehabilitation ENvironment

Participants: Ten transtibial amputees and 9 able-bodied controls

Interventions: Medio-lateral (ML) translations of the walking surface were imposed to manipulate gait stability. To provoke an adaptive gait pattern, a gait adaptability task (GA-task) was used, in which subjects had to hit virtual targets with markers guided by their knees.

Main Outcome Measures: Walking speed, step length, step frequency, step width and selected measures of gait stability (short-term Lyapunov exponents and backward and ML margins of stability (MoS))

Results: Amputees walked slower than able-bodied people, with a lower step frequency and with wider steps. This resulted in larger ML MoS, but smaller backward MoS for amputees. In response to the balance perturbation both groups decreased step length, and increased step frequency and step width. Walking speed did not change significantly in response to the perturbation. These adaptations induced an increase in ML and backward MoS. To perform the GA-task, both groups decreased step length and increased step width, but did not change step frequency and walking speed. ML and backward MoS were maintained in both groups.

Conclusion: Transtibial amputees have the capacity to use the same strategies to deal with challenges of gait stability and adaptability, to the same extent as able-bodied people.

Key words: Transtibial amputees, gait stability, gait adaptability
5.2 Introduction

Falling and fear of falling form a substantial problem in people that walk with a prosthesis\textsuperscript{92-93}. Explanations for the fact that amputees fall more often than able-bodied people are that amputees are less stable during steady state walking\textsuperscript{72,77}, are less responsive to mechanical balance perturbations and have less abilities to adapt their gait pattern to environmental changes\textsuperscript{60,65}. To improve gait stability and adaptability during the rehabilitation of amputees, insight into the strategies used by amputees to optimize these aspects of walking, and how these strategies differ from strategies selected by able-bodied people is required.

In a previous study, we found that able-bodied people, who were confronted with continuous balance perturbations during walking, decreased step length, increased step width and step frequency, and kept walking speed constant in order to maintain gait stability\textsuperscript{46}. These adaptations caused an increase in the medio-lateral and backward margins of stability (MoS), which seems to be a compensation for the decrease in local dynamic stability (LDS), caused by the perturbations. When able-bodied subjects additionally had to adapt their gait pattern to suddenly appearing cues; these people were able to maintain sufficient MoS, despite the disturbing effect of both manipulations. They did so by decreasing step length and increasing step width, but without an increase in step frequency and therefore with a decrease in walking speed. The absence of an increase in step frequency in the latter task could be explained by the fact that this would decrease the available time to respond to the presented cues\textsuperscript{47}.

The overall gait pattern of people with a lower limb amputation differs from that utilized by able-bodied people. Amputees often walk with wider steps than able-bodied controls, which can be explained as a strategy to decrease the risk of falling, because wider steps cause an increase of the MoS in medio-lateral (ML) direction\textsuperscript{23,58}. The increase in ML MoS possibly compensates for the lower LDS found for amputees, compared to able-bodied people\textsuperscript{77}. Besides, a lower preferred walking speed, with a lower cadence and smaller step length were found for amputees\textsuperscript{23,99}. However, whether these adaptations really serve the purpose of limiting fall risk or possible other purposes such as minimizing the energy cost to compensate for the mechanical constraints for propulsion\textsuperscript{64,125} has not been elucidated.

Moreover, it is unclear whether people with a lower limb amputation really do or can select strategies similar to able-bodied people when gait stability or adaptability is challenged.

The purpose of this study was to investigate which strategies transtibial amputees use to cope with challenges of gait stability and adaptability. This was done by using respectively medio-lateral balance perturbations and a gait adaptability task (GA-task), in which subjects had to hit virtual targets that were projected on a 2D screen, using virtual markers controlled
by knee motion. The purpose of this task was to simulate a real-life situation that requires accurate and fast adaptations of the normal stable gait pattern, with a limited response time, for example to avoid an obstacle that suddenly appears. Subjects walked on a self-paced treadmill, which made it possible for them to continuously adapt their walking speed. The effects of the manipulations on walking speed, step frequency, step length, and step width were measured, to investigate which strategies amputees used and how these strategies differed from strategies used by able-bodied people.

5.3 Methods

5.3.1 Subjects
Ten adult subjects with an unilateral transtibial prosthesis (age 38.8 ± 14.6 years, height 1.83 ± 0.11 m, mass 87.1 ± 10.3 kg, 9 male) and nine age-matched control subjects (age 37 ± 11.4 years, height 1.73 ± 0.08 m, mass 70.2 ± 10 kg, 4 male) participated in this study. The side of amputation was equally divided between left and right. Amputees and able-bodied controls were respectively recruited from the patient population and the employees of the Military Rehabilitation Center Aardenburg, Doorn, The Netherlands. Further characteristics of the amputees are reported in table 5.1. All amputees used their own prosthesis (six amputees walked with a fixed foot and four amputees with a flexible foot) and were able to walk in daily life without any walking device for at least 30 minutes. A minimum score of E on the SIGAM scale was necessary to participate in this study. This study was approved by the medical ethical committee (Ref: NL35402.029.11) and all subjects gave their written informed consent in accordance with university policy.
5.3.2 Equipment

All subjects walked in the Computer Assisted Rehabilitation Environment (CAREN, Motek Medical b.v., Amsterdam, The Netherlands). The CAREN system consists of an instrumented treadmill mounted onto a 6-degree-of-freedom motion platform in combination with a Virtual Environment (VE) (figure 3.1a). Twelve high resolution infra-red cameras (Vicon, Oxford, UK) were used to capture kinematic data of 16 reflective markers attached to pelvis and the lower extremities (lower body plug-in-gait)\(^{25, 68}\). The treadmill was used in the self-paced mode, which allowed subjects to modify walking speed at will. This was done by servo-controlling the motor with a real-time algorithm that took into account the pelvis position in the AP-direction, as measured by the markers attached to the pelvis, and a reference position on the treadmill, corresponding to the AP-midline of the treadmill. A safety harness system suspended overhead prevented the subjects from falling, but did not provide weight support.

<table>
<thead>
<tr>
<th>Subject #</th>
<th>Cause amputation</th>
<th>Time since amputation (months)</th>
<th>Foot</th>
<th>Company</th>
<th>Socket fitting</th>
</tr>
</thead>
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<tr>
<td>1</td>
<td>Complex regional pain syndrome</td>
<td>144</td>
<td>Axtion</td>
<td>Otto Bock</td>
<td>TSB</td>
</tr>
<tr>
<td>2</td>
<td>Traffic</td>
<td>48</td>
<td>Elite VT</td>
<td>Endolite</td>
<td>PTB*</td>
</tr>
<tr>
<td>3</td>
<td>Traffic</td>
<td>96</td>
<td>1C40</td>
<td>Otto Bock</td>
<td>PTB*</td>
</tr>
<tr>
<td>4</td>
<td>Trauma</td>
<td>24</td>
<td>Variflex EVO</td>
<td>Ossur</td>
<td>PTB*</td>
</tr>
<tr>
<td>5</td>
<td>Blast</td>
<td>27</td>
<td>Variflex EVO</td>
<td>Ossur</td>
<td>PTB*</td>
</tr>
<tr>
<td>6</td>
<td>Blast</td>
<td>17</td>
<td>Fusion</td>
<td>Ohio Willow Wood</td>
<td>PTB*</td>
</tr>
<tr>
<td>7</td>
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<td>12</td>
<td>Celsus</td>
<td>College Park</td>
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<tr>
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<td>Otto Bock</td>
<td>PTB*</td>
</tr>
<tr>
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<td>Propiofoot</td>
<td>Ossur</td>
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</tr>
<tr>
<td>10</td>
<td>Blast</td>
<td>34</td>
<td>1C40</td>
<td>Otto Bock</td>
<td>PTB*</td>
</tr>
</tbody>
</table>

TSB=Total Surface Bearing  
PTB=Patella Tendon Bearing  
*pin-fixation

**Table 5.1:** Subject characteristics for amputees
5.3.3 Protocol

Warming up

Before the protocol started, subjects performed 5 warming-up trials of 3 minutes each, to become familiar with walking on a (self-paced) treadmill, the VE and the various manipulations.

Experimental trials

The actual protocol consisted of 3 trials of 4 minutes walking at self-paced walking speed: 1) a trial of unperturbed walking, 2) a trial with continuous balance perturbations, and 3) a trial with a gait adaptability task. These trials were offered in a random order.

For the balance perturbations, translations of the walking surface in ML-direction were used, following a multi-sine function (figure 3.1c):

\[ D(t) = 0.05 \left[ 1.0 \sin(0.16 \cdot 2\pi t) + 0.8 \sin(0.21 \cdot 2\pi t) + 1.4 \sin(0.24 \cdot 2\pi t) + 0.5 \sin(0.49 \cdot 2\pi t) \right] \]

where \( D(t) \) is the translation distance (m) and \( t \) is time (s) \(^{46,85-86}\).

For the gait adaptability task (GA task) the VE was used to project targets on the screen (figure 3.1b). In addition, a projection of the markers attached to the knees was shown on the screen. Subjects were instructed to hit the targets on the screen with the projected knee markers, as close as possible to the centre of the targets. A reason to choose for this task instead of a virtual obstacle avoidance tasks is the impossibility to step over virtual objects that are projected on a 2D screen. Stepping over virtual objects would require a 3D environment. Another advantage of the adaptability task used in the current study is the possibility to quantify the performance on the task in terms of accuracy of the knee movement, while for an obstacle avoidance task only a pass and a hit can be distinguished from each other to quantify the performance. In each trial, a total of 32 targets appeared. Targets appeared at initial contact and disappeared after the duration of one gait-cycle. The positions of the targets differed randomly in side (left or right), height (120% or 140% of lower leg length), and ML-position (120% or 140% of distance between the left and right anterior superior iliac spines from the midline of the treadmill) \(^{47}\).

5.3.4 Data collection

To calculate walking speed, the speed of the treadmill was recorded. Kinematic data of markers attached to the lateral malleoli of the ankles, the pelvis (left and right anterior superior iliac spines (LASI & RASI), and left and right posterior superior iliac spines (LPSI & RPSI)), and the lateral epicondyles of the knees were collected with the Vicon system. The sample rate of data collection was 120 samples/s. The final three minutes of each trial were
used for data analysis. Before data analysis, both speed data and kinematic data, were low-pass filtered with a 4th order bi-directional Butterworth filter with a cut-off frequency of 10 Hz. However, this was not done for the calculation of the local dynamic stability (LDS), given the difficulties associated with filtering in the calculation of LDS\textsuperscript{91}.

5.3.5 Data analysis

Walking speed

Walking speed was calculated as the average treadmill speed over the final 3 minutes of each trial.

Step parameters

Step frequency was determined as the inverse of the average duration between two subsequent heel-strikes, where heel-strikes were detected as the local maxima of the position of the ankle markers in the AP-direction. Step width was calculated as ML-distance between both ankle markers at the instant of heel-contact and step length was defined as the AP-distance between these markers at the instant of heel-contact.

Gait stability

To quantify gait stability, LDS, and the margins of stability (MoS) in medio-lateral (ML) and backward (BW) direction were calculated.

LDS, a concept derived from non-linear dynamics, can be quantified by the short-term (over 0–1 step) Lyapunov exponent ($\lambda_{\text{step}}$), and is a measure for the attenuation of small perturbations that naturally occur during walking\textsuperscript{28}. The $\lambda_{\text{step}}$ expresses the logarithmic rate of divergence after a small disturbance of nearby orbits in a state space constructed from the markers placed on LASI, RASI, LPSI, and RPSI\textsuperscript{46}. Negative exponents indicate local stability and positive exponents indicate local instability, with larger exponents indicating greater sensitivity to local perturbations.

The MoS were calculated, following a method derived from the method introduced by Hof\textsuperscript{57-58}, as the difference in ML and BW direction between the extrapolated centre of mass (XCoM) and the margin of the base of support (BoS) (figure 3.2). The XCoM is a concept that takes both the position and the velocity of the centre mass (CoM) into account. Although basically similar, our method differs from Hof\textsuperscript{57-58} who used force plate data for calculating the trajectory of the CoM and the XCoM. In the current study the average of the pelvis markers was used to estimate the position of the CoM. The markers attached to the ankles were used to define the margin of the BoS. Besides, in our definition of the BW MoS, the equation XCoM-BoS was used, instead of the original equation BoS-XCoM by Hof et al.\textsuperscript{57}.
Gait adaptability

Gait adaptability was quantified by the performance on the GA-task. This performance is defined as the minimum distance between knee and target centre. For the period in which the target was visible on the screen the minimal Euclidean distances between the knee markers and the centre of the target in the plane of projection of the VE was assessed for each projected target (figure 3.1b). The average of these distances was taken to get an outcome measure for the accuracy of the knee movements in performing the gait adaptability task.

5.3.6 Statistical design

To measure the effects of the balance perturbation or the GA-task on step length, step frequency, step width, walking speed, $\lambda_{\text{step}}$, and MoS, and to investigate whether these effects differ between amputees and healthy controls, 2 × 3 factorial ANOVAs were performed. The three conditions (normal walking, perturbed walking and walking with GA-task) were used as within factor and group as between factor. Simple contrasts were used to determine for which condition (perturbed walking or walking with GA-task) the variable concerned differed from normal walking. A paired samples t-test was performed to investigate whether the performance on the GA-task differed between both groups. Statistical analyses were performed using SPSS 17.0 (SPSS Inc., Chicago, IL).

5.4 Results

One of the participating amputees (no. 7) was not able to complete the trials with the perturbation and the GA-task. Therefore the analyses were done for 9 amputees and 9 healthy controls. The data of these remaining subjects did not contain any missing values. Figure 5.1 shows the averages and standard deviations for walking speed, step frequency, step length and step width for both groups and all three conditions. Amputees walked on average slower than healthy controls (F=7.468; p=0.015; df=1). Step frequency was significantly lower in the amputee group compared to the healthy group (F=11.427; p<0.01; df=1), but step length did not differ significantly between groups (F=1.564; p=0.229; df=1). Step width was larger for the amputees than for the healthy controls (F=7.503; p=0.015; df=1). Values for $\lambda_{\text{step}}$ were significantly higher for amputees than for healthy controls, which means that the amputees were locally less stable than the able-bodied controls (F=5.476; p=0.033; df=1) (Figure 5.2a). BW MoS were smaller for amputees than for able-bodied people (F=6.728; p=0.020; df=1) (figure 5.2b). In contrast, amputees walked with larger ML MoS (F=7.774; p=0.013; df=1) (figure 5.2c).
In response to the mechanical balance perturbations both groups increased step frequency (F=26.078; p<0.01; df=1) and step width (F=37.028; p<0.01; df=1), and decreased step length (F=23.223; p<0.01; df=1), while walking speed did not change significantly (F=0.046; p=0.832; df=1). For both groups λx-step increased (F=16.886; p<0.01; df=1). Besides, ML (F=22.190; p<0.01; df=1) and BW MoS (F=21.151; p<0.01; df=1) increased for both groups in response to the perturbations. No significant group by perturbation interactions were found. During the trials with the GA-task, both groups decreased step length (F=7.177; p=0.016; df=1) and increased step width (F=85.967; p<0.01; df=1), but step frequency (F=2.004; p=0.176; df=1) and walking speed did not change significantly (F=2.203; p=0.157; df=1). Both ML (F=0.023; p=0.881; df=1) and BW MoS (F=2.425; p=0.139; df=1) were not affected significantly by the GA-task. Also for the GA-task, none of the measured variables showed a significant task by group interaction. The average distance between knee and target, as outcome measure of the performance on the GA-task, was 3.71 (+/-1.30) cm for the amputees and 3.78 (+/-1.06) cm for the able-bodied people, which was not significantly different from each other (t=0.220; p=0.829; df=16).
Figure 5.1: Average and standard deviation of walking speed (A), Step frequency (B), Step length (C), and step width (D) for amputees (white symbols; n = 9) and healthy controls (black symbols; n = 9). NW: normal walking; P: perturbation; GA: gait adaptability task. Significant group effects are indicated with +. Significant contrasts between normal walking and perturbed walking and/or between normal walking and walking with GA-task are indicated with *.
5.5 Discussion

The purpose of this study was to investigate which strategies transtibial amputees use to cope with challenges of gait stability and gait adaptability, and how these strategies differ from those used by able-bodied controls. Despite differences in gait pattern during unperturbed walking, both groups responded in a similar fashion to the manipulations of stability and adaptability. Both groups increased step frequency and step width, decreased step length, and kept walking speed constant in response to the balance perturbation. To perform the GA-task, both groups decreased step length, but did not change step frequency and walking speed significantly. In addition, an increase in step width was found in both groups. These results are in agreement with the results of previous studies in which gait stability and adaptability were manipulated in able-bodied people^{46-47,85}.

In response to the balance perturbations, local dynamic stability (LDS) decreased for both amputees and able-bodied people, as reflected by the increase of $\lambda_s$-step, but this decrease was compensated by an increase of the ML MoS and the BW MoS in backward direction, which is a direct consequence of the adaptations in the different spatio-temporal gait parameters. An increase in BW MoS directly implies a decrease in the forward MoS. However, as was found...
previously, decreasing the risk of a backward fall appears to be preferred above decreasing the risk of a forward fall when stability of walking is challenged\textsuperscript{46, 90}. To prevent a backward fall the XCoM should always be in front of the dorsal border of the BoS\textsuperscript{38, 98}. As the model of Pai and Patton\textsuperscript{98} shows, the decrease in step length, in combination with the unchanged walking speed found in the present study, caused the increase in the BW MoS in response to the perturbations. In ML-direction, subjects should prevent that the XCoM exceeds the lateral border of the BoS. Hof et al.\textsuperscript{58} demonstrated that increasing step width and step frequency contribute to an increase in ML MoS. These adaptations were found in both groups in this study.

In situations that require fast adaptations of the gait pattern in response to environmental cues, it is important that these adaptations are adequate without losing balance\textsuperscript{21, 107-108}. The results showed that both amputees and able-bodied controls were capable to maintain the ML and BW MoS during the GA-task, despite the disturbing effect of this task. To maintain the MoS, subjects increased step width and decreased step length. In contrast with the response to the balance perturbation, subjects did not increase step frequency to perform the GA-task. In all likelihood, this was because an increase in step frequency would decrease the available time to respond to the targets, which would have had a negative influence on the accuracy of the hitting movement of the knee\textsuperscript{41}.

Although the response to the manipulations of gait stability and adaptability did not differ between amputees and healthy controls, indicating similar capacity and strategy to cope with perturbations and environmental cues, there were some overall differences between both groups. It is of interest to know whether these differences might be explained as a strategy to enhance stability and adaptability in people with an amputation, or serve other purposes. The LDS for amputees was lower compared to the able-bodied people, as reflected in the higher $\lambda_{\text{step}}$ values for the amputee group. The larger step width for amputees, which resulted in a larger ML MoS, might be a strategy to compensate for this. On the other hand, the BW MoS were overall smaller for amputees than for able-bodied people, which could be taken to imply that amputees were unable to increase the BW MoS and consequently ran a larger risk of making a backward fall. However, this is unlikely, because the amputees, just as able-bodied controls, increased their BW MoS when balance was perturbed. The smaller BW MoS could therefore be considered as a detrimental effect of the overall lower walking speed, caused by the lower step frequency utilized by amputees\textsuperscript{38, 98}. This lower step frequency could be an adequate adaptation to the adaptability manipulation, but this does not explain why the amputees also walked with a lower step frequency during the normal walking condition. Plausible reasons for the overall lower step frequency, and thus the lower walking
speed, are the limited push-off power and the higher energy demands of walking with a prosthesis compared to normal walking\textsuperscript{64, 125}.

An important limitation is the small number of subjects that were included in this study, which might have affected the power of the study. Besides, when interpreting the results of the present study, it has to be taken into account that the transtibial amputees that participated in this study were all relatively young and generally good walkers. All amputees, except one, were amputated following trauma. For this group of amputees, the overall walking ability is in general higher than for people with an amputation due to vascular disorders\textsuperscript{125}. All these aspects may explain that the amputees in this study could adapt their gait pattern relatively well to the applied manipulations, but generalization to less proficient walkers needs to be done with caution. Another limitation of this study is the estimation of the CoM as the average of the markers attached to the LASI, RASI, LPSI, and RPSI of the pelvis to calculate the XCoM and subsequently the MoS in ML and BW direction. This is not the real representation of the CoM, but errors made were likely similar for both groups across conditions. Therefore, these errors would not affect differences in MoS between both groups and between conditions.

The clinical implication of this study entails that not all differences in the gait pattern between people with a lower limb prosthesis and able-bodied people are dedicated to enhancing margins of stability, but might serve other functional constraints. Nevertheless, people with a lower limb amputation can and do use effective stepping strategies to enhance margins of stability when confronted with challenges to gait stability and adaptability. Because of the relatively high walking capabilities of the participating amputees, the observations made in this study can be used as a reference in assessments of walking ability of people with a transtibial amputation in more general, as well as in defining targets for the training of adequate strategies to enhance stability and adaptability.

In conclusion, in response to the balance perturbations both transtibial amputees and able-bodied people increased their ML and BW MoS by increasing step frequency and step width, and decreasing step length, while walking speed did not change with respect to normal walking. To enhance gait adaptability, both groups increased step width and decreased step length to maintain ML and BW MoS, while step frequency was not adapted to prevent a deterioration of the accuracy of the hitting movement. During unperturbed walking, amputees walked with wider steps to increase the ML MoS, possibly to compensate for a lower LDS. Other deviations of the general gait pattern of transtibial amputees compared to able-bodied people, like the overall lower step frequency and walking speed, do not seem to
contribute to enhance stability or adaptability and should therefore be attributed to other goals or deficits in prosthetic gait.

Conflicts of interest statement
Authors state that no conflicts of interest are present in the research.

Acknowledgements
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CHAPTER 6

Stepping strategies used by post-stroke individuals to maintain margins of stability during walking

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6.1 Abstract

Background
People recovering from a stroke are less stable during walking compared to able-bodied controls. The purpose of this study was to examine whether and how post-stroke individuals adapt their steady-state gait pattern to maintain or increase their margins of stability during walking, and to examine how these strategies differ from strategies employed by able-bodied people.

Methods
Ten post-stroke individuals and 9 age-matched able-bodied individuals walked on the Computer Assisted Rehabilitation Environment. Medio-lateral translations of the walking surface were imposed to manipulate gait stability. To provoke gait adaptations, a gait adaptability task was used, in which subjects occasionally had to hit a virtual target with their knees. We measured medio-lateral and backward margins of stability, and the associated gait parameters walking speed, step length, step frequency, and step width.

Findings
Post-stroke participants showed similar medio-lateral margins of stability as able-bodied people in all conditions. This was accomplished by a larger step width and a relatively high step frequency. Post-stroke participants walked overall slower and decreased walking speed and step length even further in response to both manipulations compared to able-bodied participants, resulting in a tendency towards overall smaller backward margins of stability, and significantly smaller backward margins of stability during the gait adaptability task.

Interpretation
Post-stroke individuals have more difficulties regulating their walking speed and the underlying parameters step frequency and step length, compared to able-bodied controls. These quantities are important in regulating the size of the backward margins of stability when walking in complex environments.
6.2 Introduction

People who are recovering from a stroke have an increased risk of falling during walking\textsuperscript{124}. In the literature several causes for this increased risk of falling are suggested, such as an enlarged body sway in the frontal plane during steady state walking\textsuperscript{26, 118}, and a limited capacity to adapt the gait pattern in response to environmental demands, for example to avoid an obstacle\textsuperscript{27, 107-108}. Especially when obstacles suddenly appear and fast and accurate adaptations are necessary, the failure rate in post-stroke individuals is higher compared to able-bodied people\textsuperscript{27}. Besides, not only the higher probability of an obstacle collision, but also an impaired postural stability during and after obstacle crossing might increase fall risk in post-stroke individuals\textsuperscript{108}.

Fall risk during walking can be assessed by determining the margin of stability (MoS). The MoS is defined as the distance between the extrapolated centre of mass (XCoM) and the limits of the base of support, in which the XCoM is a concept that takes both the position and the velocity of the centre of mass (CoM) into account\textsuperscript{57}. The MoS can be calculated in both medio-lateral (ML)\textsuperscript{50, 88, 90} and antero-posterior (AP)\textsuperscript{39, 88, 90, 98} direction, in which the AP MoS is usually calculated with respect to the base of support (BoS) of the leading foot at initial contact. The difficulty with the interpretation of the AP MoS is that an increase in AP MoS in backward direction by definition implies a decrease of the AP MoS in forward direction. However, from previous experiments we know that, when balance is threatened, people prioritize an increase in backward (BW) MoS, limiting the chance of a backward loss of balance, above an increase in forward MoS\textsuperscript{7-8, 46, 90}. MoS can be regulated effectively by adjusting step parameters. From studies of Hof et al.\textsuperscript{57-59}, it appeared that possible strategies to increase the ML MoS during walking are an increase in step width and step frequency, while Espy et al.\textsuperscript{38-39} have shown that a decrease in step length and an increase in walking speed have a positive effect on the size of the BW MoS.

To assess the risk of falling, ML and BW MoS can be measured during unperturbed walking. However, measuring the MoS during more challenging walking conditions allows one to investigate whether subjects are able to use active adjustments of the gait pattern to increase or at least maintain the ML and BW MoS. Previous studies have found that able-bodied people, but also people who walk with a trans-tibial prosthesis successfully exploit such strategies. In response to continuous platform perturbations they increase step width and step frequency, resulting in an increase in ML MoS, and they decrease step length while keeping walking speed constant, resulting in an increase in BW MoS\textsuperscript{46, 48, 90}. In other recent studies we investigated whether able-bodied people and trans-tibial amputees were able to control their MoS during a task in which besides maintaining gait stability, fast and accurate
adaptations of the gait pattern had to be made, to hit virtual targets with the knees. The available response time was very short, because targets appeared within the same stride as they had to be hit. We found that both subject groups decreased step length and increased in step width in their average gait pattern. These adaptations appeared to be mainly an anticipatory strategy to facilitate the fast and accurate response necessary to hit the targets, and to prevent a loss of balance while performing the task. Simultaneously, no increase in step frequency was found in this situation, probably to prevent a further decrease of the available response time which would hamper an accurate adaptation.

During unperturbed walking, the gait pattern of post-stroke individuals already differs from the gait pattern utilized by able-bodied people and some aspects of this deviant gait pattern have been explained as mechanisms to regulate gait stability. In the study of Chen et al., the larger step width in post-stroke individuals was explained as a compensation for the larger body sway in the frontal plane. A lower walking speed in people with gait impairments is frequently explained as a strategy to increase gait stability. However, a lower walking speed may decrease the BW MoS. Besides, when a reduced walking speed coincides with a decrease in step frequency it may also have a negative effect on the size of the ML MoS. Hence, it is unknown whether and how changes in the steady state gait pattern of people who have suffered from stroke affect their MoS and whether people after stroke can adapt their steady state gait pattern to increase or preserve their MoS during challenging walking conditions. Therefore in the current study we manipulated gait stability and gait adaptability during walking. We assessed whether post-stroke individuals use similar strategies as able-bodied people to preserve MoS during unperturbed walking and when required to withstand manipulations of gait stability or to facilitate gait adaptability. We hypothesized that post-stroke individuals walk with smaller MoS, compared to the able-bodied controls, and that MoS decreased even further, for the post-stroke individuals, during the manipulations of gait stability and adaptability. The main reason for these differences in MoS between both groups might be the lower walking speed, which will influence the size of the BW MoS negatively, and the lower step frequency which will decrease the ML MoS.
6.3 Methods

6.3.1 Subjects

Ten adult subjects who had suffered from a stroke (age 60.8 +/- 8.4 years, height 1.79 +/- 0.07 m., mass 88.4 +/- 8.5 kg.) and 9 age-matched control subjects (age 57.3 +/- 7.2 years, height 1.77 +/- 0.08 m., mass 79.7 +/- 9.0 kg.) participated in this study. Post-stroke participants and able-bodied controls were respectively recruited from the patient population and the employees of the Military Rehabilitation Center Aardenburg, Doorn, The Netherlands. A minimum score of 4 on the Functional Ambulation Categories (FAC)\textsuperscript{61}, in combination with a minimum score of 45 on the Berg Balance Scale (BBS)\textsuperscript{113} was required to participate in this study. Further characteristics of the post-stroke group are reported in table 6.1. This study was approved by the medical ethical committee (Ref: NL35402.029.11) and all subjects gave their written informed consent in accordance with university policy.

Table 6.1: Characteristics stroke post-stroke individuals

<table>
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<tr>
<th>Subject#</th>
<th>Cognitive disturbances</th>
<th>Side hemiparesis</th>
<th>Berg Balance Scale</th>
<th>Time since stroke (months)</th>
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<td>None</td>
<td>left</td>
<td>56</td>
<td>8</td>
</tr>
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<td>2</td>
<td>Minor</td>
<td>left</td>
<td>51</td>
<td>10</td>
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<tr>
<td>3</td>
<td>Aphasia</td>
<td>left</td>
<td>52</td>
<td>8</td>
</tr>
<tr>
<td>4</td>
<td>None</td>
<td>left</td>
<td>51</td>
<td>4</td>
</tr>
<tr>
<td>5*</td>
<td>None</td>
<td>right</td>
<td>47</td>
<td>9</td>
</tr>
<tr>
<td>6</td>
<td>Moderate</td>
<td>left</td>
<td>47</td>
<td>38</td>
</tr>
<tr>
<td>7</td>
<td>None</td>
<td>right</td>
<td>52</td>
<td>1</td>
</tr>
<tr>
<td>8</td>
<td>Aphasia</td>
<td>right</td>
<td>56</td>
<td>1</td>
</tr>
<tr>
<td>9</td>
<td>None</td>
<td>right</td>
<td>45</td>
<td>1</td>
</tr>
<tr>
<td>10</td>
<td>Minor</td>
<td>left</td>
<td>55</td>
<td>12</td>
</tr>
</tbody>
</table>

*Excluded from data analyses
6.3.2 Equipment

All subjects walked in the Computer Assisted Rehabilitation (CAREN, Motek Medical b.v., Amsterdam, The Netherlands), which consists of an instrumented treadmill mounted onto a 6-degree-of-freedom motion platform in combination with a Virtual Environment (VE) (figure 3.1a). Twelve high resolution infra-red cameras (Vicon, Oxford, UK) were used to capture kinematic data of 16 reflective markers attached to pelvis and the lower extremities (lower body plug-in-gait\textsuperscript{25,68}). The treadmill was used in the self-paced mode, which allowed subjects to modify walking speed at will. This was done by servo-controlling the motor with a real-time algorithm that took into account the pelvis position in the AP-direction, as measured by the markers attached to the pelvis, and a reference position on the treadmill, corresponding to the AP-midline of the treadmill. A safety harness system suspended overhead prevented the subjects from falling, but did not provide weight support.

6.3.3 Protocol

Familiarization

Before the protocol started, subjects performed at least 5 familiarization trials of 3 minutes each, to become familiar with walking on a (self-paced) treadmill, the VE and the various manipulations.

Experimental trials

The actual protocol consisted of 3 trials of 4 minutes walking at self-paced walking speed: 1) a trial of unperturbed walking, 2) a trial with a continuous perturbation of the motion platform, and 3) a trial with a gait adaptability task. The first minute of each trial was used to let subjects get used to the self-paced setting of the treadmill and the manipulation concerned. All trials were offered in random order.

For the platform perturbation, translations of the walking surface in ML-direction were used, following a multi-sine function\textsuperscript{46,85-86} (figure 3.1c).

For the gait adaptability task (GA task) the VE was used to project targets on the screen (figure 3.1b). In addition, a projection of the markers attached to the knees was shown on the screen. Subjects were instructed to hit the targets on the screen with the projected knee markers that were attached to the lateral epicondyles, as close as possible to the centre of the targets. The purpose of this task was to simulate a situation that requires accurate and fast adaptations of the normal stable gait pattern, with a limited response time, for example to avoid an obstacle that suddenly appears. A reason to choose for this specific task instead of a virtual obstacle avoidance tasks is the impossibility to step over virtual objects that are projected on a 2D screen. Stepping over virtual objects would require a 3D environment. Another advantage of the adaptability task used in the current study is the possibility to
quantify the performance on the task in terms of accuracy of the knee movement, while for an obstacle avoidance task only a pass and a hit can be distinguished from each other to quantify the performance. Within this trial, a total of 32 targets appeared with a time interval of about 5 seconds in between. Targets appeared at initial contact and disappeared after approximately one gait-cycle, the duration of which was estimated in the first minute of the trial. The positions of the targets differed randomly in side (left or right), height (120% or 140% of lower leg length), and ML-position (120% or 140% of distance between the left and right anterior superior iliac spines from the midline of the treadmill), to increase the unpredictability of this task.

6.3.4 Data collection
Kinematic data of markers attached to the lateral malleoli of the ankles, the pelvis (left and right anterior superior iliac spines (LASI & RASI), and left and right posterior superior iliac spines (LPSI & RPSI)), and the lateral epicondyles of the knees were collected with the Vicon system. The sample rate of data collection was 120 samples/s. The final three minutes of each trial were used for data analysis. Before data analysis, both speed data and kinematic data, were low-pass filtered with a 4th order bi-directional Butterworth filter with a cut-off frequency of 10 Hz.

6.3.5 Data analysis
All outcome measures, walking speed, the different step parameters and the margins of stability, were averaged across the total numbers of steps during the final 3 minutes of each trial. However, before calculating these outcome measures for the trials with GA-task, we removed the strides in which the targets had to be hit, to focus on the anticipatory strategy to facilitate the gait adaptations.

Walking speed
Walking speed was calculated as the average treadmill speed over the final 3 minutes of each trial.

Step parameters
Step frequency was determined as the inverse of the average duration between two subsequent heel-strikes, where heel-strikes were detected as the local maxima of the position of the ankle markers in the AP-direction. Step width was calculated as ML-distance between both ankle markers at the instant of heel-contact and step length was defined as the AP-distance between these markers at the instant of heel-contact.

Gait stability
To quantify gait stability, margins of stability (MoS) in medio-lateral (ML) and backward (BW) direction were calculated, following a method derived from the method introduced by
Hof et al.\textsuperscript{57-58}, as the difference in ML and BW direction between the extrapolated centre of mass (XCoM) and the margin of the base of support (BoS) (figure 3.2). The XCoM is a concept that takes both the position and the velocity of the centre mass (CoM) into account. MoS were calculated for the instant at which the MoS reached its minimum value within each step, which is always at the instant of initial contact for the BW MoS.

Our method is basically similar to that of Hof \textsuperscript{57-58} who used force plate data for calculating the trajectory of the CoM and the XCoM. The difference with the current study is that the average of the pelvis markers was used to estimate the position of the CoM. The markers attached to the ankles were used to define the margin of the BoS. Besides, in our definition of the BW MoS, the equation XCoM-BoS was used, instead of the original equation BoS-XCoM by Hof et al.\textsuperscript{57}.

To quantify the amount of body sway in the frontal plane the maximal difference in XCoM between subsequent steps was calculated (ML XCoM\textsubscript{disp}; figure 3.2a). This method adds to the method used in previous studies \textsuperscript{26,118}, in which only the displacement of the CoM was used to calculate the amount of body sway during walking. By taking the XCoM also the velocity of the CoM was taken into account.

To differentiate between the contribution of step length and walking speed on the size to the BW MoS, we additionally calculated the distance between the BoS and the CoM in backward direction (BW BoS-CoM\textsubscript{dist}; figure 3.2b) and the forward velocity of the CoM (FW vCoM) at initial contact. Resulting values were averaged over steps.

\textit{Gait adaptability}

Gait adaptability was quantified by the performance on the GA-task. This performance is defined as the minimum distance between knee and target centre. For the period in which the target was visible on the screen the minimal Euclidean distances between the knee markers and the centre of the target in the plane of projection of the VE was assessed for each projected target. The average of these distances was calculated.

\textbf{6.3.6 Statistical design}

To determine the effects of the platform perturbation and the GA-task on step length, step frequency, step width, walking speed, ML XCoM\textsubscript{disp}, and ML and BW MoS, and to investigate whether these effects differ between post-stroke participants and able-bodied people, 2 × 3 factorial ANOVAs were performed. The three conditions (normal walking, perturbed walking and walking with GA-task) were used as within factor and group as between factor. Simple contrasts were used to determine whether outcome measures differed between normal walking and either perturbation task or adaptability task, and whether there were group by condition interaction effects. P-values less than 0.05 were considered significant. In the case
of a significant interaction effect, paired-samples t-tests with a Bonferroni correction (critical p-value: 0.025) were performed to investigate for each group separately whether the parameter concerned was affected by the manipulation. A Mann-Whitney U test was used to investigate whether the performance on the GA-task differed between groups. For this comparison, a non-parametric test was chosen because the distribution of the performance scores was not normal for the post-stroke group. Also for the outcome of the Mann-Whitney U test, a p-value less than 0.05 was considered significant Statistical analyses were performed using IBM SPSS Statistics 20.0.

6.4 Results
One of the participating post-stroke participants (subject number 5, table 6.1) was not able to complete the trials with the perturbation and the GA-task. Therefore, the analyses were achieved in nine post-stroke participants and nine able-bodied controls. A visual representation of the results for the gait parameters (walking speed, step frequency, step length and step width) and the margin of stability measures is presented in figures 6.1, 6.2 and 6.3. Results of the statistical analyses are shown in table 6.2.

A significantly smaller step length, a decreased walking speed, and a larger step width were observed in the post-stroke group compared to the controls. Step frequency did not differ significantly between both groups. Also for the ML and BW MoS no significant group effects were found, although the difference in BW MoS between groups nearly reached the level of significance (p=0.077), with smaller margins for the post-stroke participants. The XCoM$_{disp}$ was significantly larger and BW BoS-CoM$_{dist}$ and FW vCoM were significantly smaller for the post-stroke group compared to the able-bodied controls.

A main effect of the platform perturbation on all recorded gait parameters was found. Subjects increased step width and step frequency and decreased step length and walking speed in response to the perturbation. For step length and walking speed, group × perturbation interactions were found. Post hoc analyses showed that step length decreased in both groups, but this decrease was larger in the post-stroke group. Walking speed decreased only in the post-stroke group. In response to the GA-task, step length and walking speed decreased, step width increased, while step frequency was not significantly affected by the GA-task. For walking speed, a group × GA-task interaction effect was found. Post hoc analyses showed that in response to the GA-task walking speed decreased only in the post-stroke group.

In response to the platform perturbation, both ML XCoM$_{disp}$ and the ML MoS increased in both groups. BW MoS was not affected significantly, while BW BoS-CoM$_{dist}$ and FW vCoM
decreased in both groups. For FW vCoM and BW BoS-CoM\textsubscript{dist}, also significant group ×
perturbation interactions were found. Post-hoc test showed that FW vCoM only decreased
for the post-stroke group (p<0.01 for the post-stroke group; p=0.204 for the able-bodied
controls). BW BoS-CoM\textsubscript{dist} decreased in both groups (p<0.01 for both groups), but this
decrease was larger in the post-stroke group.

In response to the GA-task, neither the BW MoS nor the ML MoS was affected significantly.
However, a significant group × GA-task interaction effect for the BW MoS was found. Post-
hoc analyses showed that BW MoS significantly increased in response to the GA-task in the
able-bodied participants (p<0.01), while BW MoS slightly decreased in the post-stroke group,
however this was not significant (p=0.104). ML XCoM\textsubscript{disp} increased significantly, while BW
BoS-CoM\textsubscript{dist} and FW vCoM decreased significantly in response to the GA-task. For both BW
BoS-CoM\textsubscript{dist} and FW vCoM also a significant group × GA-task interaction effect was found.
Post-hoc analyses showed that both BW BoS-CoM\textsubscript{dist} and FW vCoM only decreased in the
post-stroke group. The performance on the GA-task, quantified as the average distance
between knee and target, was 13.02 (+/-18.04) cm in the post-stroke participants and 3.74
(+/-1.02) cm in the able-bodied participants, and differed significantly between both groups
(U=77.00; p<0.001; df=16).

Chapter 6
Table 6.2: Results for statistical analyses

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Perturbation contrast</th>
<th>GA-task contrast</th>
<th>Main effect group</th>
<th>Group×perturbation contrast</th>
<th>Group×GA-task contrast</th>
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<td>Step length</td>
<td>F = 50.675</td>
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<td>F = 11.300</td>
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<td>Step frequency</td>
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<td>ML MoS</td>
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<td>p &lt; 0.01</td>
<td>p &lt; 0.01</td>
<td>p = 0.409</td>
</tr>
<tr>
<td></td>
<td>df=1,16</td>
<td>df=1,16</td>
<td>df=1,16</td>
<td>df=1,16</td>
<td>df=1,16</td>
</tr>
</tbody>
</table>

*Significant at the 0.05 level
Figure 6.1: Average and standard deviation of walking speed (A), Step frequency (B), Step length (C), and step width (D) for post-stroke individuals (n = 9) and healthy controls (n = 9). NW: normal walking; P: perturbation; GA: gait adaptability task. Significant group effects are indicated with ‡. Significant contrasts between normal walking and perturbed walking and/or between normal walking and walking with GA-task are indicated with * and significant group × manipulation interaction effects are indicated with +.

Figure 6.2: Average and standard deviation of ML MoS (A) and ML XCoM_{disp} (B) for post-stroke individuals (n = 9) and healthy controls (n = 9). NW: normal walking; P: perturbation; GA: gait adaptability task. Significant group effects are indicated with ‡. Significant contrasts between normal walking and perturbed walking and/or between normal walking and walking with GA-task are indicated with *. For these data no significant group × manipulation interaction were found.
Figure 6.3: Average and standard deviation of BW MoS (A) BW BoS-CoM dist (B) and FW vCoM (C) for post-stroke individuals (n = 9) and healthy controls (n = 9). NW: normal walking; P: perturbation; GA: gait adaptability task. Significant group effects are indicated with ‡. Significant contrasts between normal walking and perturbed walking and/or between normal walking and walking with GA-task are indicated with * and significant group × manipulation interaction effects are indicated with +.

6.5 Discussion

The purpose of the current study was to evaluate whether post-stroke individuals preserve their margins of stability during walking with manipulations of gait stability and gait adaptability and whether post-stroke individuals use similar strategies to withstand perturbations of gait stability or to facilitate gait adaptability as able-bodied people. For the trials of unperturbed walking and for walking with the platform perturbation we looked at the average adaptations of the gait pattern. For the trial with the GA-task we were interested in the anticipatory adaptations in the steady state gait pattern to facilitate the fast and accurate response necessary to hit the targets, and therefore we removed the strides in which the targets had to be hit. Post-stroke participants walked with comparable ML MoS compared to able-bodied participants, and were able to increase their ML MoS to the same degree as able-bodied participants in response to both manipulations of gait stability and gait adaptability, which was in contrast with our hypothesis. With respect to the BW MoS, these margins were smaller compared to the able-bodied group. This was especially the case for the condition with the GA-task, which appeared to be a challenging task for the post-
stroke participants given both a smaller BW MoS and a reduced accuracy of the knee movements compared to the able-bodied controls, which was in line with our hypothesis. Although post-stroke participants did not actually fall during the experiment, both reduced backward margins of stability and reduced movement accuracy would bring post-stroke individuals at a higher risk of disrupting forward progression and possibly falling in situations during normal live walking in which fast and accurate adaptations of gait pattern are required.

In the current study, we found an overall larger excursion of the ML XCoM in the post-stroke group compared to the able-bodied controls, indicating larger ML body sway. In concert with that we found that post-stroke individuals walked with a larger step width and, considering their relatively low walking speed, a relatively high step frequency, compared to able-bodied participants. Both increases in step width and frequency have been shown to enhance ML MoS. Therefore it seems valid to conclude that these gait adaptations allow post-stroke individuals to maintain a ML MoS of the same size, compared to able-bodied subjects, despite the larger excursion of the ML XCoM in this group. It could be argued that the larger step width might be the cause of the larger body sway instead of a compensatory strategy to deal with this larger sway. However, from the mechanical model introduced by Hof it can be derived that an increase in step width will result in a net increase of the ML MoS despite a concomitant increase in body sway. Therefore increasing step width should likely be regarded as a strategy to deal with the consequences of increased ML sway and not only the cause of this increased sway.

For both groups ML excursions of the XCoM were larger during both the manipulations of gait stability and adaptability. In response to the platform perturbation post-stroke participants were able to withstand this manipulation, by increasing their ML MoS to the same degree as able-bodied participants, by further increasing step width and step frequency. Similarly, for the condition with the GA-task, the regulation of the ML MoS did not differ between both groups. Both groups were able to maintain their ML MoS, despite the disturbing effect of the manipulation, by increasing their step width. In agreement with earlier studies an increase in step frequency was absent in both groups, probably to prevent a decrease in available response time, which might affect the accuracy of the knee movement needed to hit the targets. These findings of perturbed and unperturbed walking corroborate the notion that increasing step width and step frequency are functional gait adaptations that can be and are used by post-stroke individuals to preserve ML MoS.

In contrast with the regulation of the ML MoS, differences in the regulation of the BW MoS were observed between post-stroke participants and able-bodied participants. The BW MoS
tended to be smaller for post-stroke participants compared to controls, especially for the trial in which the GA-task had to be performed. In response to the GA-task, able-bodied participants increased their BW MoS by reducing step length without a concomitant reduction in walking speed. Post-stroke participants also did reduce step length in response to the GA-task, but this was accompanied by a reduction of walking speed. BW MoS can be increased by either decreasing step length (reducing the distance between CoM and BoS at initial contact) or by increasing forward velocity at initial contact. The results shown in figure 6.3 demonstrate that the distance between the CoM and the leading foot at initial contact was indeed smaller for post-stroke participants as a result of the smaller step length. However, forward velocity of the CoM at initial contact was considerably lower compared to able-bodied controls, especially during the trial with the GA-task. In contrast to the general notion that a lower walking speed might be a strategy to increase gait stability, these results demonstrate that the lower walking speed in post-stroke individuals seems to cause a decrease in the BW MoS, and might therefore increase the risk of a disruption of forward progression and possibly a backward fall. It remains speculative why post-stroke individuals have problems with the regulation of the BW MoS. Possibly, post-stroke individuals experience problems with selecting an appropriate combination of step length and frequency, at a given walking speed, as evidenced by the seemingly inappropriate reduction in step length and speed in response to the GA-task. This lack of adaptation might have a physical cause (i.e. impaired movement selectivity or a reduced push-off capacity of the hemiparetic leg) or a mental cause (a higher fear of falling or a conflict between the cognitive demands of the GA-task and walking ability). From the data in the current study this cannot be resolved.

A limitation of this study is the estimation of the CoM as the average of the markers attached to the LASI, RASI, LPSI, and RPSI of the pelvis to calculate the XCoM. It is conceivable that the post-stroke group has a larger sway of the upper body, in which case this method could cause an underestimation of the displacement of the XCoM, and therefore an overestimation of the size of the MoS. However, from a study of De Bujunda et al. it appeared that, at least in the frontal plane, displacements and accelerations of the shoulders and pelvis during walking are very similar, and this was the case for both post-stroke individuals and able-bodied controls. Therefore, we expect that the potential error made in the estimation of the CoM position in the current study will be small. Besides, this error likely affects only the overall group difference in the MoS, and not the adaptation of these measures in response to the manipulations.
When interpreting the results of the present study, it has to be taken into account that the post-stroke individuals that participated in this study were all relatively good walkers. This may explain why the post-stroke subjects were able to walk with comparable ML MoS as the able-bodied controls and why all subjects, except one, were able to complete the protocol without actually falling. Therefore generalization of the results to subjects with a more severe hemiparesis needs to be done with caution.

Secondly, it is of importance to mention that the calculation of the MoS as a measure of stability is based on an inverted pendulum model, which is a strong simplification of human walking. This model is designed to quantify the contribution of a change in foot placement, like an increase in step width, in the regulation of dynamic balance. However, wider steps will also result in larger angular and linear momenta of the pendulum at foot contact. This is trivial for an inverted pendulum, because its legs are rigid, but in humans larger linear and angular momenta at foot placement require larger joint moments in the ‘new’ stance leg to reduce and reverse these momenta. This aspect of balance control is ignored in the model. The results of the current study have shown that the relatively large step width in post-stroke individuals result in a proper foot placement with respect to the XCoM. However, generating sufficient joint moments to compensate for the increased momenta, might be a problem in post-stroke individuals, especially in view of associated muscle weakness. Controlling the size of the MoS is just one prerequisite for preventing falls during walking, and consequently the translation of the results for the MoS to the more general concept of fall risk, should be done with care.

In conclusion, despite the larger ML body sway, post-stroke participants were able to regulate their ML stability, during all experimental conditions, to the same degree as the able-bodied subjects. However this required a larger step width and a relatively high step frequency compared to able-bodied participants. BW MoS tended to be smaller for post-stroke participants, compared to able-bodied participants, especially for the condition in which the GA-task had to be performed. In response to the GA-task, BW MoS decreased for the post-stroke group, while able-bodied participants were able to maintain their BW MoS. An explanation for these smaller BW MoS seems to be that post-stroke participants significantly decreased their walking speed, while able-bodied participants maintained their walking speed in response to both manipulations though an effective adaptation of both step length and step frequency. Future studies should aim on identifying whether post-stroke individuals are really limited in selecting different step length and frequency combinations at a constant walking speed and which impairments cause these limitations.
Conflicts of interest statement
Authors state that no conflicts of interest are present in the research.

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Stepping asymmetry among individuals with unilateral transtibial limb loss could be functional in terms of gait stability

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Physical Therapy; Accepted for publication.
7.1 Abstract

Background
The asymmetry in step length in prosthetic gait is often seen as a detrimental effect of the impairment, however this asymmetry might also be a functional compensation. An advantage of a smaller non-prosthetic step length, and specifically foot forward placement (FFP), might be that it will bring the centre of mass closer to the base of support of the leading foot, and therefore increases the backward margin of stability (BW MoS).

Objective
Investigating whether difference in step length and FFP between prosthetic and non-prosthetic step are compatible with a strategy to regulate the BW MoS.

Design
Observational and cross-sectional study

Methods
Ten people after transtibial amputation walked for 4 minutes on a self-paced treadmill. Step length and FFP were calculated at initial contact. The size of the BW MoS was calculated for the moment of initial contact and at the end of the double-support phase.

Results
Step length (5.4%;p=0.030) and FFP (7.9%;p=0.015) were shorter for the non-prosthetic step than for the prosthetic step. The BW MoS at initial contact was larger for the non-prosthetic step (p=0.007), but because of a significant leg by gait event interaction effect (p=0.008) MoS did not differ significantly at the end of the double-support phase.

Limitation
Participating subjects were relatively good walkers (SIGAM E).

Conclusion
The smaller step length and FFP of the non-prosthetic step help to create a larger BW MoS at initial contact for the non-prosthetic step compared to the prosthetic step. Step length asymmetry in people after transtibial amputation might hence be seen as a functional compensation to preserve BW MoS during double support to cope with the limited push off power of the prosthetic ankle.
7.2 Introduction

One of the characteristics of the gait pattern of people after transtibial amputation, is an asymmetry in step length. Although the direction of the asymmetry in step length might vary across individuals, on group level, the step of the non-prosthetic leg, in which the healthy leg is the leading leg, is generally found to be shorter compared to the step with the prosthetic leg\textsuperscript{4, 67, 84, 131}. A commonly used explanation for a shorter step with the non-prosthetic leg is the reduced push off capacity of the prosthetic leg\textsuperscript{64, 102, 131}. Besides, prosthetic misalignment, stump pain or discomfort might contribute to step length asymmetry\textsuperscript{1, 22}.

Gait asymmetry in people after transtibial amputation is often seen as a negative consequence of the disorder. However, although asymmetry may be detrimental from a cosmetic perspective, there is no evidence that step length asymmetry is necessarily detrimental from a functional point of view. In fact, a shorter non-prosthetic step could even be beneficial. Modeling studies and experimental studies with able-bodied controls and people after transtibial amputation\textsuperscript{33-34, 64}, suggest that a shorter (non-prosthetic) step length might limit the metabolic cost of the step-to-step transition during the double support phase. Another possible benefit of a shorter non-prosthetic step length, which to our knowledge has not been considered explicitly before, could be the regulation of stability during walking.

A basic requirement of walking is that the CoM passes the dorsal border of the stance foot during each single support phase, otherwise forward progression will be interrupted and a backward fall might even occur in case a recovery response, like stepping back, fails. In studies of Pai and Patton\textsuperscript{98} and Espy et al.\textsuperscript{38-39} the risk on such a backward loss of balance, was quantified as the distance between the kinematic state (position and velocity) of the centre of mass (CoM) and the dorsal border of the base of support (BoS) of the leading foot (figure 7.1), in the current study designated as the backward margin of stability (BW MoS). Hof et al.\textsuperscript{57, 59} provided an analytical expression for this concept in which this CoM state is quantified as the extrapolated centre of mass (XCoM). Following this concept, increasing the forward velocity of the centre of mass (vCoM) and/or decreasing step length (and more specifically the foot forward placement (FFP), defined as the distance leading foot and the CoM) will increase the BW MoS. A larger BW MoS implies that it is easier for the CoM to pass the posterior border of the BoS defined by the new stance leg, during the following single support phase and will decrease the risk of a backward loss of balance.
From previous studies on the regulation of gait stability it appeared that able-bodied subjects, but also people after transtibial amputation, maintained walking speed and decreased step length in response to continuous balance perturbations, which resulted in an increase in BW MoS. However, these studies focused on average changes in step length and BW MoS. It is unknown whether and how differences in step length between sides influence the BW MoS for the affected and unaffected step. Possibly, shortening the non-prosthetic step length can be regarded a proper strategy to increase the BW MoS for this step. The reduced push-off capacity of the prosthetic leg often results in a limited capacity to deliver mechanical power with the prosthetic leg during the push-off. This might result in a lower vCoM, and therefore a smaller BW MoS, during the step-to-step transition from prosthetic to non-prosthetic leg. A shorter non-prosthetic step might therefore serve the purpose to compensate for this detrimental effect of lower vCoM on the BW MoS, as it brings the leading foot closer to the CoM at initial contact, which increases BW MoS. Note, however, that this will only be true when this asymmetry in step length will
manifest itself in an asymmetry in FFP, and not in the distance between the trailing foot and the CoM (in this study called the trunk progression (TP)), because the latter will not influence the size of the BW MoS (figure 7.1)\textsuperscript{98}.

The purpose of this observational study was to characterize differences in step length, and more specific in FFP, between prosthetic and non-prosthetic steps of people after transtibial amputation and the concomitant difference in BW MoS between these steps. We hypothesized that people after transtibial amputation walk with a shorter intact step length and shorter FFP of the intact step. We expected that vCoM will be reduced during the step-to-step transition from prosthetic to intact leg, as a result of the lack of push-off capacity of the prosthetic leg. Finally, we hypothesized that a smaller FFP for the non-prosthetic step will compensate for the reduction in vCoM to preserve BW MoS.

### 7.3 Methods

#### 7.3.1 Subjects

Ten adult subjects with a unilateral transtibial prosthesis participated in this observational study. Subjects were recruited from the patient population of the National Military Rehabilitation Center Aardenburg, Doorn, The Netherlands. This study was approved by the medical ethical committee (Ref: NL35402.029.11) and all subjects gave their written informed consent in accordance with university policy.

#### 7.3.2 Equipment

During the experiment the Computer Assisted Rehabilitation Environment (CAREN, Motek Medical b.v., Amsterdam, The Netherlands) was used. Subjects walked on a treadmill within a virtual environment to create an optical flow during the walking trials. Twelve high resolution infra-red cameras (Vicon, Oxford, UK) were used to capture kinematic data of 16 reflective markers attached to pelvis and the lower extremities (lower body plug-in-gait). The treadmill was used in the self-paced mode, which allowed subjects to walk at their comfortable walking speed and allowed spontaneous variations in walking speed. This self-paced mode was used to approach over ground walking more closely, while recording a large number of steps within a single trial. In the self-paced mode, the motor of the treadmill was servo-controlled by a real-time algorithm that took into account the pelvis position in the anterio-posterior (AP) direction, as measured by the markers attached to the pelvis, and a reference position on the treadmill, corresponding to the AP-midline of the treadmill. A safety harness system suspended overhead prevented the subjects from falling, but did not provide weight support.
7.3.3 Protocol
Before the experimental trial, subjects performed 5 warming-up trials of 3 minutes each to become familiar with walking on a (self-paced) treadmill and the virtual environment. The experimental trial consisted of 4 minutes walking at self-paced walking speed. During the first 30 seconds of this trial, subjects walked at a fixed walking speed, which was substantially lower than their comfortable walking speed, determined during one of the warming-up trials. After 30 seconds the self-paced mode of the treadmill was started after which the subjects could walk at their comfortable walking speed for the remaining 3.5 minutes.

7.3.4 Data collection
Data collection took place for the final three minutes of the experimental trial. The speed of the treadmill was recorded and kinematic data of markers attached at the lateral malleoli of the ankles, the pelvis (left and right anterior superior iliac spines (LASI & RASI), and left and right posterior superior iliac spines (LPSI & RPSI)) were collected with the Vicon system. The sample rate of data collection was 120 samples/s. Data were low-pass filtered with a 4th order bi-directional Butterworth filter with a cut-off frequency of 10 Hz.

7.3.5 Data analysis
Step length, foot forward placement, and trunk progression
Step length was defined as the AP-distance between the ankle markers at the instant of heel-contact, where heel-contacts were detected as the local maxima of the position of the ankle markers in the AP-direction. The body’s centre of mass (CoM) position was approximated by the centre of the polygon described by the four markers attached to the pelvis. Foot forward placement (FFP) and trunk progression (TP) were respectively calculated as the fore-aft distance between the ankle marker of the leading foot and the CoM and the ankle marker of the trailing foot and the CoM, at initial contact. These definitions for FFP and TP were derived from studies of Roerdink et al. However, we have chosen to use a slightly different definition for TP, to be able to evaluate asymmetries in FFP and TP independently from each other. In the studies of Roerdink et al, TP was defined as the distance traveled by the centre of mass (CoM) between subsequent steps, which is partly determined by the length of the FFP of the previous step. Asymmetry indices were calculated for step length, FFP and TP as: (value NP/(( value NP + value P)/2)) * 100%, in which NP stands for the non-prosthetic step and P for the prosthetic step.

Velocity centre of mass
The velocity of the CoM (vCoM) was calculated as the first derivative of the position of the centre of the pelvis relative to the global reference frame plus the velocity of the treadmill.
This was done for both initial contact and the moment of toe-off of the contralateral leg. Moments of toe-off were detected as the local minima of the position of the ankle markers in the AP-direction.

**Backward margins of stability**

To calculate the backward margin of stability (BW MoS), a method derived from the procedure introduced by Hof⁵⁷ was used. In the current study, the extrapolated centre of mass (XCoM), was estimated by taking the CoM position plus its velocity times a factor \( \sqrt{\frac{l}{g}} \), with \( l \) being the maximal height of the estimated CoM and \( g \) the acceleration of gravity. The MoS was calculated as the position of the XCoM minus the position of the lateral malleolus of the ankle of the leading foot (representing the border of the base of support) for the moment of initial contact and the moment of toe-off of the contralateral foot in anterio-posterior direction. Although basically similar, our method differs from that of Hof⁵⁷ who used force plate data instead of kinematic data for calculating the XCoM and MoS.

**7.3.6 Statistical design**

To test whether differences between legs (non-prosthetic and prosthetic) in step length, FFP and TP at initial contact were significant, paired samples t-tests were used. To investigate whether BW MoS and vCoM differed between prosthetic and non-prosthetic steps (factor: leg) and whether these variables changed during double support, i.e. between initial contact and toe-off of the contra lateral foot (factor: gait event), \( 2 \times 2 \) factorial ANOVAs were performed with the two legs and the two gait events as within variables. P-values less than 0.05 were considered significant. When an interaction effect was present paired-samples t-tests with a Bonferroni correction (critical p-value: 0.0125) were used to investigate for each leg separately whether the considered variable changed between both gait events and to test for each gait event separately whether there was a difference between legs. Statistical analyses were performed using SPSS 17.0 (SPSS Inc., Chicago, IL).

**7.4 Results**

Characteristics of the people after transtibial amputation who participated in the study are reported in table 5.1. The side of amputation was equally divided between left and right. All subjects used their own prosthesis and were able to walk in daily life without any walking device for at least 30 minutes. All were classified with a minimum score of E on the SIGAM scale (Able to walk more than 50 meters. Independent of walking aids, except occasionally for confidence or to improve confidence in adverse terrain or weather)¹⁰⁵. All subjects were able to complete the walking trial on the treadmill and walked on average at a speed of 1.22 m/s (SD: 0.22). Step length and FFP were significantly smaller
(respectively 5.4 and 7.9 percent) for the non-prosthetic step compared to the prosthetic step, while TP did not differ between legs (figure 7.2 and table 7.1). Asymmetry indices for step length, FFP and TP were respectively 97.2, 95.9 and 98.9.

Figure 7.3 shows the BW MoS (upper panel) and vCoM (lower panel) at initial contact and subsequent contralateral toe off of the prosthetic and non-prosthetic step. There were significant main effects of leg and gait event on the size of the MoS, with a larger MoS for the non-prosthetic step and an increase of the MoS between initial contact and contralateral toe-off. In addition, there was a significant leg by gait event interaction effect (table 7.1). Post-hoc analyses (table 7.2) showed that MoS increased for both legs, but this increase was larger between initial contact and contralateral toe off for the prosthetic step compared to the non-prosthetic step. Besides, post-hoc analyses also showed that the difference in MoS between legs was only present for the moment of initial contact.

No significant main effects of leg and gait event on vCoM were found, but there was a significant leg by gait event interaction effect (table 7.1). Post-hoc tests (table 7.2) revealed a decrease of vCoM during the double-support phase of the non-prosthetic step, while vCoM did not change significantly for the prosthetic step. At initial contact there was a trend towards a higher vCoM for the non-prosthetic step compared to the prosthetic step. At contralateral toe-off vCoM was smaller for the non-prosthetic step, but this did not reach the level of significance either.
Figure 7.2: The average and standard deviation ($n=10$) of the prosthetic (O) and non-prosthetic ($\Delta$) step length (SL), foot forward placement (FFP) and trunk progression (TP) at initial contact.

Figure 7.3: The average and standard deviation ($n=10$) of the backward margin of stability (MoS; upper panel) and the velocity of the centre of mass ($v_{CoM}$; lower panel) at initial contact (IC) and contra-lateral toe-off (TO) of the prosthetic (O) and non-prosthetic ($\Delta$) step.
Table 7.1: Results for the primary statistics

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Main effect leg(^1)</th>
<th>Main effect gait event(^2)</th>
<th>Leg × gait event interaction effect</th>
</tr>
</thead>
<tbody>
<tr>
<td>Step length</td>
<td>t=2.581; (p=0.030*); df=9</td>
<td>N.A.</td>
<td>N.A.</td>
</tr>
<tr>
<td>FFP</td>
<td>t=3.006; (p=0.015*); df=9</td>
<td>N.A.</td>
<td>N.A.</td>
</tr>
<tr>
<td>TP</td>
<td>t=0.490; (p=0.636); df=9</td>
<td>N.A.</td>
<td>N.A.</td>
</tr>
<tr>
<td>BW MoS</td>
<td>t=3.723; (p=0.005*); df=9</td>
<td>t=0.883; (p=0.400); df=9</td>
<td>t=18.151; (p&lt;0.001*); df=9</td>
</tr>
<tr>
<td>vCoM</td>
<td>t=3.823; (p=0.005*); df=9</td>
<td>t=0.883; (p=0.400); df=9</td>
<td>t=18.151; (p&lt;0.001*); df=9</td>
</tr>
</tbody>
</table>

\(^1\)Difference between prosthetic and non-prosthetic leg.
\(^2\)Difference between initial contact and contralateral toe-off.

* Significant at the 0.05 level

Table 7.2: Results for the post-hoc statistics

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Between legs - at IC</th>
<th>Between legs - at contralateral TO</th>
<th>Between IC and contralateral TO - prosthetic leg</th>
<th>Between IC and contralateral TO - non-prosthetic leg</th>
</tr>
</thead>
<tbody>
<tr>
<td>BW MoS</td>
<td>t=3.723; (p=0.005*); df=9</td>
<td>t=0.883; (p=0.400); df=9</td>
<td>t=18.151; (p&lt;0.001*); df=9</td>
<td>t=8.875; (p&lt;0.001*); df=9</td>
</tr>
<tr>
<td>vCoM</td>
<td>t=3.823; (p=0.005*); df=9</td>
<td>t=0.883; (p=0.400); df=9</td>
<td>t=18.151; (p&lt;0.001*); df=9</td>
<td>t=8.875; (p&lt;0.001*); df=9</td>
</tr>
</tbody>
</table>

* Significant at the 0.0125 level
7.5 Discussion

The purpose of the current study was to investigate whether differences in step length between prosthetic and non-prosthetic steps, and more specifically in foot forward placement (FFP), in prosthetic gait are compatible with a strategy to regulate gait stability in terms of the backward margin of stability (BW MoS). Step length was on average shorter for the non-prosthetic step, which is the step in which the healthy leg was leading. A remarkable result is that the step length asymmetry found in the current study was entirely due to an asymmetry in FFP. At initial contact the BW MoS was larger for the non-prosthetic step compared to the prosthetic step, but this difference had disappeared at the end of the double support phase. The average velocity of the centre of mass (vCoM) did not differ between steps, but in contrast to the prosthetic step, vCoM decreased significantly during double-support following the non-prosthetic step.

The results found in the current study support our hypothesis that a shorter non-prosthetic step length in people after transtibial amputation contributes to a larger BW MoS at initial contact of the non-prosthetic step. This seems to compensate for the limited capacity of people after transtibial amputation to regulate the size of the BW MoS during the double-support phase following the non-prosthetic step. During this double support phase the lack of ankle push-off of the prosthetic leg causes a decrease in the vCoM, which limits the increase of the BW MoS during double support. A smaller non-prosthetic FFP at initial contact may be necessary to compensate for this limited increase in BW MoS during the following double support phase, and as such decrease the risk of interruption of forward progression and possibly even falling backward after contralateral toe off. In figure 7.4, a schematic overview is given of the effect of the decrease of the non-prosthetic step length on the BW MoS, based on the results of the current study (figure 7.4 II), compared to the BW MoS of the prosthetic step (figure 7.4 I) and compared to the possible consequence of an absence of a shorter non-prosthetic step length on the BW MoS (figure 7.4 III).

The clinical implication of the present results is that retaining a symmetrical step length during the rehabilitation of people after transtibial amputation might result in difficulties with the regulation BW MoS during walking. Gait symmetry should therefore not be a primary goal in rehabilitation of people after amputation. Pursuing step length symmetry seems to be justified only when sufficient push off power in the prosthetic leg can be guaranteed to prevent a drop in vCoM during the gait cycle. With current prostheses, people compensate for a lack of ankle push off by generating positive work through hip extension prior to initial contact or by using a ‘controlled falling’ strategy in which potential energy is transformed into kinetic energy. However, this seems insufficient to preserve
similar BW MoS during the double support phase compared to the contralateral step. To restore gait symmetry while preserving BW MoS, the push off capacity of prosthetic feet should be further enhanced. Currently advanced dynamic feet are under development that might fulfill this purpose\(^{53, 109}\). Whether the risk of a backward loss of balance is indeed a problem in walking with a lower limb prosthesis can be debated. However, the results of our previous study indicate that people after transtibial amputation prefer to increase their BW MoS above increasing their forward MoS when balance mediolateral perturbations were imposed during walking\(^{48}\). This could be explained by the fact that a backward loss of balance requires a reversal of the periodic leg movement (stepping backward) to regain balance, while a forward loss of balance requires a relatively small adaption of the next step(s), like a temporary increase in step length, to recover\(^{59}\). Besides, in contrast to a forward loss of balance, a backward loss of balance automatically implies that forward progression will be interrupted.

When interpreting the results of the present study, it has to be taken into account that the people after transtibial amputation that participated in this study were all relatively young and generally good walkers (minimum score of E on the SIGAM\(^{105}\)). All subjects were able to walk without a device in daily life for at least 30 minutes consecutively, and walked at a relatively high walking speed (in comparison an average walking speed of 1.43 m/s was found in a group of age-matched controls while walking on the self-paced treadmill\(^{48}\)). All people after transtibial amputation, except two, were amputated following trauma. In this group, the overall walking ability is in general higher than in people after amputation due to vascular disorders\(^{126}\). The results of this study might also be applicable for people after transfemoral amputation or people with other unilateral impairments, for example caused by a stroke, however generalization of the results need to be confirmed in these specific groups. Secondly, it should be noted that there was a lot of heterogeneity in the energy storage and return capacities of the prosthetic feet used by the subjects. Despite this heterogeneity we have found structural differences between the prosthetic and the non-prosthetic step. However, it can be presumed from our results that increasing push off capacity of the prosthetic foot and ankle will reduce the drop in CoM velocity during the step to step transition and preserve BW MoS without the need for a shorter step. Because of the relatively small sample size we could not perform a valid post hoc comparison of the effect of foot type. A future analysis of the effect of foot type will be of interest.

The conclusions drawn in the current study are based on an observation of prosthetic gait. Therefore, it remains difficult to conclude whether the regulation of the BW MoS is the primary reason for the asymmetry in step length. There might be multiple other explanations.
for gait asymmetry in people after transtibial amputation, like the minimization of the metabolic cost during the step to step transition\textsuperscript{64} and pain or an incorrect alignment of the prosthesis\textsuperscript{1,22}. Therefore, the apparent effects of a shorter non-prosthetic step length on the MoS could also be a coincidental side effect. To further support our hypothesis that step length asymmetry enhances BW MoS, an experimental approach should be taken. For instance the use of manipulations that force people after transtibial amputation to walk symmetrically or a comparison between walking with solid versus dynamic feet (that partly restore push off capacity) might be used to establish a causal effect between step length differences and BW MoS.

In conclusion, step length asymmetry with a smaller FFP of the non-prosthetic step compared to the prosthetic step can be observed in people after transtibial amputation. This asymmetry preserved the BW MoS at the end of double support after the non-prosthetic step, in spite of a reduction of CoM velocity during double support. Consequently, step length asymmetry in prosthetic gait could potentially be regarded as a functional adjustment to preserve sufficient BW MoS and prevent a backward fall in the presence of limited push-off ability of the prosthetic leg. The results of this study illustrate that the asymmetry in gait pattern for people after transtibial amputation is not necessarily a detrimental effect of the impairment, but could be beneficial in the regulation of gait stability. This should be considered in gait training for people after transtibial amputation.
Figure 7.4: Schematic overview of possibilities to regulate the BW MoS of the non-prosthetic step (II and III) compared to the prosthetic step (I). For the prosthetic step the BW MoS increases during the double support phase as a result of a change in CoM state during this phase (Ia versus Ib). IIa: A shorter non-prosthetic step due to a smaller FFP, results in a larger BW MoS at initial contact (compared to prosthetic step in I). IIb: A decrease of vCoM during the double-support phase of the non-prosthetic step causes a smaller increase of the BW MoS compared to the prosthetic resulting in a BW MoS that is about the same size as the BW MoS at the end of the double support phase of the prosthetic step (compared to prosthetic step in Ia). IIIa: A non-prosthetic step with the same step length as the prosthetic step results in a BW MoS of comparable size as for the prosthetic step at initial contact (I). IIIb: A decrease of vCoM during the double-support phase of the non-prosthetic step causes a smaller increase of the BW MoS compared to the prosthetic step resulting in a smaller BW MoS compared to the BW MoS at the end of the double support phase of the prosthetic step (compared to prosthetic step in Ia).
Conflicts of interest statement
Authors state that no conflicts of interest are present in the research.

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CHAPTER 8

Summary and general discussion
8.1 Introduction
The general aim of this thesis was to investigate which gait strategies can be used to decrease fall risk during walking and whether people with gait impairments are able to use these strategies. In the first chapters, experiments were presented with able-bodied subjects. The results of these experiments helped us to understand which gait strategies can be used, in the absence of any impairments, to withstand manipulations of gait stability (chapter 2) and gait adaptability (chapter 3), and how these adjustments influence the magnitude of the short-term Lyapunov exponent ($\lambda_s$) and the size of the margins of stability (MoS; chapter 4). In chapters 5 and 6, studies were presented in which the aim was to investigate whether people with a unilateral gait impairment, caused by a transtibial amputation (chapter 5) or a stroke (chapter 6), are able to use comparable strategies to withstand manipulations of gait stability and adaptability. Finally, in chapter 7, the possible role of step length asymmetry was examined in transtibial amputees in the regulation of backward (BW) MoS.

8.2 Gait adjustments used by able-bodied people to regulate gait stability and adaptability
In this paragraph an answer will be given on the first two research questions of this thesis:
1. Which gait strategies do able-bodied people use to cope with manipulations of gait stability and gait adaptability?
2. How do manipulations of stride frequency, stride length and walking speed affect the short-term Lyapunov exponent and the margins of stability in able-bodied people?
In chapters 2 and 3, balance perturbations and a gait adaptability task (GA-task) were used to assess how subjects adjusted their gait pattern, and whether gait stability changed, in response to these manipulations. In these chapters gait stability was quantified by $\lambda_s$ (chapter 2) and the medio-lateral (ML) and BW MoS (chapters 2 and 3). In chapter 4, we manipulated stride frequency, stride length and walking speed, to investigate how these adjustments influenced both the $\lambda_s$ and the MoS.

8.2.1 Responses to the manipulation of gait stability
In chapter 2, a continuous medio-lateral translation of the walking surface following a pseudo-random pattern was used to perturb gait stability in able-bodied people (figure 2.1). In response to this perturbation, subjects decreased step length and increased step frequency and step width, while keeping walking speed constant. Besides, subjects became locally less stable, as indicated by the increase in $\lambda_s$, and subjects increased their ML and BW MoS in response to the perturbation. Following the theoretical models forming the basis for
calculating the MoS, the increase in ML MoS can be explained as a consequence of the increases in step frequency and step width\textsuperscript{58-59}, and the increase of BW MoS as a result of the decrease in step length\textsuperscript{38,98}. Based on the results of this study it was not possible to conclude whether the increase in $\lambda_S$ was solely an effect of the perturbation, or whether the gait adjustments partly counteracted this increase in $\lambda_S$.

8.2.2 Responses to the manipulation of gait adaptability

In chapter 3, a GA-task was used to investigate which strategies able-bodied people used to maintain balance while at the same time fast and accurate adaptations in the movement pattern had to be made to hit virtual targets that appeared on the screen (figure 3.1). To investigate the potential conflict between the maintenance of balance and the facilitation of accurate adaptations, we offered the GA-task both in absence and presence of the balance perturbation described in chapter 2. While performing the GA-task, both with and without perturbation, subjects decreased their step length and walking speed, maintained step frequency and slightly increased step width. Again these adaptations in the gait pattern can be explained as a strategy to preserve or enhance the ML and BW MoS. When comparing the adaptations found in response to the GA-task (chapter 3) with the adaptations found in response to the balance perturbation (chapter 2), the most notable difference is the absence of an increase in step frequency, while performing the GA-task. A possible explanation for this difference might be that an increase in step frequency would reduce the available time to respond to the targets, which might have a detrimental effect on the accuracy of the adaptation\textsuperscript{10,20,41}.

8.2.3 Responses to the manipulation of the gait pattern

To investigate whether the gait adjustments that were observed in response to manipulations of gait stability and adaptability could indeed account for the changes found in $\lambda_S$ and the ML and BW MoS, in chapter 4, step frequency, step length and walking speed were manipulated systematically and independently. Local dynamic stability, expressed as the $\lambda_S$, did not change in response to the imposed differences in the gait pattern. ML MoS increased with an increase in step frequency, while BW MoS increased with an increase in walking speed and with a decrease in step length. These results were in line with the expectations based on the behavior of a simple inverted pendulum\textsuperscript{59,98}. Although the inverted pendulum model is a strong simplification of human walking, this model appeared to be applicable to understand and predict MoS during human walking.
8.2.4 Conclusions
The adjustments in the gait pattern in response to the balance perturbation served the purpose to increase the ML and BW MoS, possibly to compensate for the decrease in local dynamic stability resulting from the perturbation. In response to the GA-task, both with and without perturbation, it appeared that able-bodied subjects were able to maintain their ML and BW MoS, while at the same time facilitating the required accuracy of the adaptation. As already mentioned in paragraph 8.1, the results of the studies performed in able-bodied controls provided a reference with respect to the strategies that can be used to increase or at least maintain gait stability during challenging walking conditions. Subsequently, it is of interest to investigate whether people with gait impairments have the ability to use the same strategies in response to the manipulations of gait stability and adaptability and whether differences in the gait pattern during unperturbed walking between these groups can be attributed to the regulation of the MoS.

8.3 Gait stability and adaptability in people with gait impairments
Based on the results of the studies presented in chapters 5 and 6, we will answer the third research question:
3. Which strategies do transtibial amputees and post-stroke individuals use to withstand manipulations of gait stability and gait adaptability and do these strategies differ from the strategies used by able-bodied controls?
In these studies we first investigated whether there were overall differences in the gait pattern and gait stability between amputees and post-stroke individuals on the one hand and able-bodied controls on the other hand, during unperturbed walking. Subsequently, the response to the manipulations of gait stability and adaptability for the different patient groups were compared with able-bodied controls. By comparing the responses to the perturbation with the differences in unperturbed walking between patients and controls, it is possible to examine whether these differences in the unperturbed gait pattern can be interpreted as a strategy to enhance gait stability.

8.3.1 Differences in the unperturbed gait pattern
Both transtibial amputees and post-stroke individuals walked overall slower compared to able-bodied controls. For transtibial amputees this was mostly due to a lower step frequency. As step length did not differ between the amputee group and the able-bodied group, the lower walking speed resulted in a lower BW MoS for transitibial amputees, in line with the relation between walking speed and the BW MoS observed in chapter 4. In contrast to the amputees, post-stroke individuals walked slower because of a reduced step length. Because
of the positive effect of shorter steps on the size of the BW MoS, the lower walking speed only caused a tendency towards lower BW MoS for post-stroke individuals compared to able-bodied controls. Both transtibial amputees and post-stroke individuals walked with wider steps. For the transtibial amputees this resulted in a larger ML MoS, despite the lower step frequency. The larger ML MoS might compensate for the reduced local dynamic stability (i.e. higher values for $\lambda_s$) found for the transtibial amputees. For post-stroke individuals the ML MoS did not differ from able-bodied controls, although post-stroke individuals not only walked with a larger step width, but also with a relatively high step frequency. This limited ML MoS arose from a larger ML body sway in the post-stroke individuals, which resulted in a larger ML amplitude of the extrapolated centre of mass (XCoM).

8.3.2 Gait adjustments in response to the balance perturbation and the GA-task
The adjustments in the gait pattern in response to the balance perturbation and the GA-task did not differ between amputees and able-bodied controls and were in line with the adaptations found in the studies described in chapter 2 and 3. For both these groups the adjustments in the gait pattern resulted in an increase in ML and BW MoS in response to the balance perturbations and maintenance of both MoS while performing the GA-task. Besides, the accuracy of knee-movements during the GA-task did not differ between both groups. In contrast to the transtibial amputees, post-stroke individuals differed in their response to the manipulations compared to the able-bodied controls. In response to both manipulations post-stroke individuals decreased their already lower walking speed even further, which would reduce the BW MoS. However in response to the platform perturbation, they also showed a relatively large decrease in step length, which resulted in BW MoS of a comparable size for both groups during the perturbation trial. In response to the GA-task, the decrease in walking speed for the post-stroke individuals was not accompanied by this relatively large decrease in step length, which caused a smaller BW MoS in the post-stroke group compared to the able-bodied group. In addition, also the accuracy of knee movements while performing the GA-task was worse for the post-stroke individuals compared to the able-bodied controls.

8.3.3 Conclusions
We concluded that, just as the able-bodied subjects, the transtibial amputees who participated in our study were able to select a strategy resulting in an increase in ML and BW MoS when stability is perturbed, and a maintenance in MoS in situations that require fast and accurate adaptations of the gait pattern while retaining balance. Because transtibial amputees were, among others, able to increase the BW MoS in response to the balance perturbation, the overall lower BW MoS, during perturbed and unperturbed walking, seems
Chapter 8

not a result of a lack of balance control in this direction. Probably it should be considered as a side effect of other constraints of walking with a prosthesis, like a reduced walking speed to compensate for the higher energetic demands of walking with a prosthesis, although this strategy is at the cost of the size of the BW MoS.

In contrast to the amputees, post-stroke individuals had difficulties with the regulation of the BW MoS, especially during the GA-task. These results could imply that post-stroke individuals lack the ability to adjust their gait pattern in such a way that MoS are preserved. Besides, the accuracy of the knee movements, necessary to hit the targets during the GA-task, was worse for the post-stroke group. This suggests that they are at an increased risk of falling, when obstacles have to be avoided during life walking, because of a higher probability of an obstacle hit, but also of a higher chance on a backward loss of balance. This is in line with previous experimental findings. Possibly, training post-stroke individuals to use adjustments of step frequency and step length, while keeping walking speed constant, might help them to better control the BW MoS. Such a training might therefore help to create a gait pattern that is more robust against perturbations that have may cause a backward fall or slip, when walking in a challenging environment. This aspect will be discussed further in paragraph 8.6.

8.4 The role of step length asymmetry in the regulation of the backward margins of stability in transtibial amputees

In the preceding paragraphs, we focused on the role of average adjustments in step parameters over both legs. However, one of the characteristics of people with an unilateral gait impairment is an asymmetry between legs during walking. For example, the step of the unaffected leg, in which the healthy leg is the leading leg, is generally found to be shorter compared to the step with the affected leg. Gait asymmetry is often seen, as a detrimental effect of a disorder, but a shorter unaffected step might be beneficial in the regulation of the BW MoS. Based on the results presented in chapter 7, we will answer the fourth research question:

4. How does step length asymmetry in amputees contribute to the regulation of the backwards margins of stability?

From this observational study, it appeared that a shorter non-prosthetic step (the step in which the healthy leg is leading), and more specifically a shorter non-prosthetic foot forward placement (distance between leading foot and centre of mass at initial contact), resulted in a larger BW MoS at initial contact compared to the prosthetic step. This large BW MoS at initial contact of the non-prosthetic step, appeared to be a compensation for a limited increase in
velocity of the centre of mass during the transition from non-prosthetic to the prosthetic step. Consequently, at the end of this transition phase, the BW MoS was sufficient to prevent a backward loss of balance. Although, it is difficult to conclude whether the regulation of the BW MoS is the primary reason for the asymmetry in step length, this observation at least illustrates that an asymmetry is not necessarily a detrimental effect of the impairment, but could be beneficial in the regulation of gait stability. Therefore, pursuing a symmetric gait pattern during rehabilitation may not be indicated.

8.5 General implications

In the previous sections of this epilogue, answers were given to the four research questions that were formulated in the general introduction of this thesis. These research questions contribute to the general aim of this thesis, namely to investigate which strategies can be used to decrease fall risk during walking and whether people with gait impairments are able to use these strategies. Measures that were used in this thesis to estimate fall risk were \( \lambda_s \) and the ML and BW MoS. The main part of the studies that have related adjustments in the gait pattern to changes in gait stability before, have focused on changes in walking speed\(^{13, 29, 31, 37, 70}\). The motivation to perform the study was often the assumption that a lower walking speed, frequently found in people with gait impairments, might be a functional compensation to increase gait stability\(^{29, 31, 37, 70}\). This assumption could however be debated based on the findings of this thesis, as will be explained in the first part of this paragraph. Secondly, we will focus on the relation between the stability measures used in this thesis and the actual fall risk.

8.5.1 Walking speed and gait stability

In contrast to the belief that a lower walking speed might be a functional compensation to increase gait stability, it appeared from the results of chapter 2 of this thesis that subjects did not decrease walking speed, but adjusted stride frequency and stride length in response to the manipulation of gait stability. In line with this finding, it might be the case that the results of the studies that have investigated the relation between walking speed and gait stability previously, were influenced by unnoticed changes in stride frequency and stride length throughout the experiment. In chapter 4, we therefore investigated whether walking speed, independently from changes in stride frequency and stride length, influenced local dynamic stability and the ML and BW MoS. As already discussed in paragraph 8.2, both changes in walking speed and changes in stride length or step frequency, did not influence the magnitude of \( \lambda_s \), but did affect the size of the ML and BW MoS. With respect to the latter result we can conclude that a decrease in walking speed will decrease the BW MoS, and will
decrease the ML MoS when the decrease in walking speed coincides with a decrease in stride frequency. Therefore, based on the results presented in this thesis, a lower walking speed appears not to be a functional adaptation to increase gait stability, but could actually be a cause of the decrease in gait stability in people with gait impairments.

However, the results described above might be specific for the conditions imposed during the experiments described in this thesis. It may be that walking speed is reduced in people with gait impairments to allow more time to anticipate to challenges of stability that can be foreseen. The gait adaptability task that was used in our studies allowed very limited response time and moreover given the task design slowing down would not substantially increase the available response time. This may be different in everyday walking situations.

### 8.5.2 Gait stability and fall risk

We calculated the $\lambda_s$ and ML and BW MoS as measures for gait stability, and more indirect as estimates of fall risk. The relation between these stability measures and the actual fall risk warrants further discussion.

In paragraph 1.2 of the general introduction we stated that the $\lambda_s$ and the MoS cannot be interpreted independently from each other in relation to the risk of falling. A small $\lambda_s$ does not automatically imply that the risk of falling is low, because simultaneously the MoS can be very small. Conversely, large MoS might not be large enough in the case of a very high $\lambda_s$ (see also figure 1.1). From the results presented in this thesis, it appears that $\lambda_s$ increases as a consequence of external perturbations and gait impairments. Interestingly, adjusting the gait pattern does not help to counteract this increase in $\lambda_s$, but serves the purpose to increase the MoS, and therefore to increase the robustness of the system (The walls of the bowl in figure 1.1 becomes less steep, and therefore the size of the bowl has to increase). However, whether this increase in MoS actually decreases the risk of falling, or whether this increase in MoS is necessary to maintain fall risk at the same level, is difficult to conclude.

A negative MoS will not necessarily result in a fall. In the case of a negative ML or BW MoS, respectively a crossing step or a backward step can be made, to prevent a fall. Able-bodied subjects walked with smaller ML MoS compared to transtibial amputees. Obviously, this does not indicate that these people are at an increased risk of falling. Instead, this may reflect that able-bodied people have less difficulties in making a recovery step in the case of a loss of balance. These subjects might prefer to walk with smaller MoS, because this is for example less energy consuming. However, although the size of the MoS might not be directly related to fall risk, when people adjust their gait pattern to increase their MoS, this indicates at least that a strategy is chosen to prevent that an unexpected perturbation will result in a fall.
In paragraph 1.2 it was mentioned that an increase in BW MoS increases the risk of a forward loss of balance. However, in contrast to a backward loss of balance, a forward loss of balance can be corrected by only a (small) adaptation of the next step(s), like a temporary increase in step length\textsuperscript{59}. Therefore, controlling the BW MoS might be more important than controlling the forward (FW) MoS. The results presented in this thesis support this assumption, because able-bodied people, but also people with a transtibial amputation spontaneously increased their BW MoS, instead of the FW MoS, in challenging walking conditions. However, it has to be noted that these results might be specific for the conditions imposed during the experiments described in this thesis. In the case of, for example, stepping off a curb\textsuperscript{3}, or walking in a condition with a high risk of a forward trip\textsuperscript{114}, increasing the FW MoS might be prioritized. In addition, it would be of interest to investigate whether walking over ground, instead of on a treadmill, influences the choice between increasing the BW or FW MoS.

8.6 Clinical implications

The studies presented in this thesis showed that people with gait impairments, like the post-stroke individuals that participated in our study, are not always able to adequately adjust the gait pattern to preserve or enhance their MoS in response to both the balance perturbation and the GA-task used. In clinical practice, this balance perturbation and GA-task, in combination with the possibility to give real-time feedback about the adjustments of the gait pattern might be used to detect difficulties in the use effective gait strategies to decrease fall risk.

From the results of the study described in chapter 6, it appeared that post-stroke individuals have difficulties with the adaptation of step frequency, step length and walking speed, to preserve their BW MoS. However, the outcome of this study did not provide information about why post-stroke individuals did not select the gait strategy that enhances or preserves the MoS in response to the manipulations. The lack of an appropriate adjustment might both have a physical cause (i.e. impaired movement selectivity or a reduced push-off capacity of the hemiparetic leg\textsuperscript{2, 94, 101, 117}) or a mental cause (a higher fear of falling\textsuperscript{82} or a conflict between the cognitive demands of the GA-task and walking ability\textsuperscript{66, 100, 111}). In this paragraph, a pilot study will be described, in which we investigated whether post-stroke individuals are physically able to adjust their gait pattern, to increase the MoS during walking.
8.6.1 The ability of post-stroke individuals to use gait strategies to increase margins of stability

Based on the results presented in chapter 6, training post-stroke individuals to select a gait pattern that results in larger MoS could be considered as an effective intervention to reduce fall risk. To investigate whether such training would be feasible in post-stroke individuals, we performed a pilot study in which we investigated whether post-stroke individuals are able to walk at different combinations of stride frequency and stride length, and secondly we investigated how these gait manipulations influenced the size of the BW MoS. Subjects walked at different walking speeds by either increasing stride length, stride frequency or both (figure 8.1). During these trials the treadmill was set at the walking speed that was required for each specific trial and subjects received visual feedback about the required and current stride length (figure 8.2). This feedback method allowed some variation in stride length, stride frequency, and walking speed during each trial.

| V = 100%  | V = 111%  | V = 123%  |
| SF = 111% | SF = 111% | SF = 111% |
| SL = 90%  | SL = 100% | SL = 111% |
| V = 100%  | V = 111%  | V = 100%  |
| SF = 100% | SF = 100% | SF = 90%  |
| SL = 100% | SL = 100% | SL = 111% |

Figure 8.1: Schematic overview of the experimental conditions (V: Walking speed; SL: Stride length, SF: Stride frequency). Percentages are percentages of respectively comfortable walking speed, stride length, and stride frequency. The rows represent the stride frequency manipulations and the columns the stride length manipulation. Note that walking speeds on the diagonals represent a constant percentage of comfortable waking speed.
In figures 8.3 and 8.4, an overview of results obtained in a group of 10 post-stroke individuals, is given. In figure 8.3 the difference between the observed average stride length and the imposed stride length (averaged over all subjects and expressed as a percentage of the required stride length) is shown.

Post-stroke individuals appeared to have the greatest difficulty executing the required task, when they needed to increase stride frequency. Increasing walking speed by means of increasing stride length was less problematic. This was evidenced by the fact that the deviation from the designated stride length was higher when post-stroke individuals were required to walk at higher stride frequencies than self-selected.

In figure 8.4, the results for the BW MoS are shown for all experimental trials. Although post-stroke individuals appeared to have difficulties with increasing their stride frequency, and are not fully able to walk at a stride frequency equal to 111% of their comfortable stride frequency, the BW MoS were higher for these trials compared to comfortable walking. This is in line with the results presented in chapter 4. Therefore, training post-stroke individuals to further increase stride frequency during walking might be beneficial to control the BW MoS. However further research is needed to investigate which underlying impairments of the post-stroke individuals caused the difficulty to increase stride frequency and whether increasing stride frequency comes at the costs relative to other constraints in walking (i.e. limiting the energy cost). Besides, it has yet to be proven whether training to adjust stride frequency and stride length at various walking speeds will also help post-stroke individuals.
to select a more effective strategy with respect to the maintenance of the MoS during challenging walking conditions.

**Figure 8.3**: Difference between the average stride length used and the stride length subjects were instructed to use, averaged over 10 subjects and expressed as a percentage of the required stride length for all experimental trials (SL = stride length; SF = stride frequency). The red bars represent the trials for which the deviation from the required stride length, significantly differed from the deviation during comfortable walking (100% stride length and 100% stride frequency; white bar). The green bars represent the trials for which the deviation from the required stride length, was not significantly different from the deviation during comfortable walking.
Figure 8.4: Backward margins of stability (BW MoS), averaged over 10 subjects, for all experimental trials (SL = stride length; SF = stride frequency). The red bars represent the trials for which the BW MoS significantly differed from the BW MoS during comfortable walking (100% stride length and 100% stride frequency; white bar). The green bars represent the trials for which the BW MoS were not significantly different from the BW MoS during comfortable walking.

8.7 General conclusions

The research presented in this thesis contributes to a better understanding of how adjustments of the gait pattern can contribute to maintain or even increase gait stability, when facing balance manipulations, or being required to make fast and accurate adaptations of the gait pattern. It was demonstrated that some of the differences in the gait pattern between people with gait impairments and able-bodied people, like for example the larger step width in both amputees and post-stroke individuals, or the asymmetry in step length for amputees, might represent a functional adaptation in view of the regulation of the size of the margins of stability, while other gait changes, such as the reduced walking speed in amputees and post-stroke individuals appear less functional.

Especially post-stroke individuals appeared to have difficulties to select a strategy that enhances or preserves backward margins of stability, in challenging walking conditions. Future research should reveal which impairments or other constraints (for example minimizing energy cost) cause this suboptimal gait pattern in terms of the MoS. Secondly, it would be of interest to investigate whether specific training on the adjustments of step frequency and step length, helps these people to enhance margins of stability in response to
balance perturbations and to conditions requiring gait adaptability and subsequently whether such a training does reduce fall incidence in daily life.
NEDERLANDSE SAMENVATTING

Stap voor Stap;

Stapstrategieën om vallen tijdens het lopen te voorkomen
Inleiding

Personen met een beperking van de loopvaardigheid, bijvoorbeeld veroorzaakt door een onderbeenamputatie of een cerebrovasculair accident (CVA) hebben een verhoogd risico op vallen tijdens het lopen, in vergelijking tot valide personen. Dit leidt naast fysieke problemen, vaak ook tot emotionele problemen zoals het ontwikkelen van angst voor vallen, wat een grote belemmering kan zijn in het dagelijks leven. Om deze reden is het van belang om meer kennis te ontwikkelen over mogelijke oorzaken van het verhoogd risico op vallen bij mensen met een beperking van de loopvaardigheid en vooral over mogelijke methodes om dit risico op vallen te verkleinen.

Het algemene doel van het in dit proefschrift gepresenteerde onderzoek was om te onderzoeken welke strategieën, in termen van aanpassingen in het looppatroon, gebruikt kunnen worden door personen met een beperking van de loopvaardigheid om het risico op vallen te verkleinen; en in hoeverre deze personen dergelijke strategieën al gebruiken. De specifieke vraagstellingen die in dit proefschrift beantwoord zijn waren:

1. Welke stapstrategieën gebruiken valide proefpersonen om manipulaties van de stabiliteit en manipulaties die een beroep doen op het adaptatievermogen te weerstaan tijdens het lopen?
2. Hoe beïnvloeden manipulaties van stapfrequentie, staplengte en loopsnelheid de ‘short-term Lyapunov exponent’ en de medio-laterale en achterwaartse stabiliteitsmarge?
3. Welke strategieën gebruiken personen met een onderbeenamputatie en personen na een CVA om manipulaties van stabiliteit en manipulaties die een beroep doen op het adaptatievermogen te weerstaan, en in hoeverre verschillen deze strategieën van de strategieën die worden gebruikt door valide personen?
4. Op welke manier draagt de asymmetrie in staplengte bij personen met een onderbeenamputatie bij aan de regulatie van de achterwaartse stabiliteitsmarge?

In de eerste hoofdstukken zijn experimenten beschreven die zijn uitgevoerd bij valide proefpersonen. De resultaten van deze experimenten hebben ons geholpen om te begrijpen welke strategieën door mensen zonder een beperking van de loopvaardigheid gebruikt (kunnen) worden om manipulaties van de stabiliteit (hoofdstuk 2) en manipulaties die een beroep doen op het adaptatievermogen (hoofdstuk 3) tijdens het lopen te weerstaan. Daarnaast hebben we in hoofdstuk 4 onderzocht hoe het gebruik van deze strategieën de ‘short-term Lyapunov exponent’ ($\lambda$) en de stabiliteitsmarges in medio-laterale en achterwaartse richting (medio-lateral and backward margins of stability (ML and BW MoS)) beïnvloedt. $\lambda$ is een maat die beschrijft hoe snel een persoon terugkeert naar de oorspronkelijke of een nieuwe evenwichtssituatie na een verstoring (zie figuur 1.2), terwijl
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de MoS kwantificeren hoe groot een verstoring kan zijn voordat er daadwerkelijk balansverlies optreedt. De MoS worden berekend als het verschil tussen het geëxtrapoleerde lichaamszwaartepunt (extrapolated centre of mass (XCoM)) en het steunvlak (base of support (BoS))(zie figuur 1.4). In hoofdstuk 5 en 6 zijn studies beschreven waarin het doel was om te onderzoeken in hoeverre personen met een beperking van de loopvaardigheid, veroorzaakt door een amputatie van het onderbeen (hoofdstuk 5) of een CVA (hoofdstuk 6) in staat zijn om manipulaties van de stabiliteit en manipulaties die een beroep doen op het adaptatievermogen te weerstaan en in hoeverre zij daarbij gebruik maken van dezelfde strategieën als valide personen. Tot slot is in hoofdstuk 7 beschreven wat de mogelijke rol is van een asymmetrie in staplengte bij personen die lopen met een onderbeenprothese in de regulatie van de BW MoS.

Stapstrategieën gebruikt door valide personen om stabiliteit en het adaptatievermogen te reguleren.

Op basis van de studies gepresenteerd in hoofdstukken 2, 3 en 4, kan een antwoord gegeven worden op de eerste twee onderzoeksvragen van dit proefschrift:

1. Welke stapstrategieën gebruiken valide proefpersonen om manipulaties van de stabiliteit en manipulaties die een beroep doen op het adaptatievermogen te weerstaan tijdens het lopen?

2. Hoe beïnvloeden manipulaties van stapfrequentie, staplengte en loopsnelheid λs en de ML en BW MoS?

In hoofdstuk 2 en 3 zijn een balansverstoring (hoofdstuk 2) en een adaptatietaak (hoofdstuk 3) gebruikt om te achterhalen hoe proefpersonen hun looppatroon aanpassen, en hoe de stabiliteit van het looppatroon verandert in reactie op deze manipulaties. De stabiliteit van het lopen is in deze hoofdstukken gekwantificeerd door λs (hoofdstuk 2) en de ML en BW MoS (hoofdstuk 2 en 3). Om en beter begrip te krijgen over de mogelijke relatie tussen de stapstrategieën gevonden in hoofdstuk 2 en 3 en de stabiliteit van het lopen hebben we in hoofdstuk 4 stapfrequentie, staplengte en loopsnelheid systematisch gemanipuleerd, om het effect van deze aanpassingen op zowel λs als de ML en BW MoS te onderzoeken.

Respons op de stabiliteitsmanipulatie

In hoofdstuk 2 is een continue zijwaartse translatie van het loopoppervlak gebruikt, volgens een pseudo-random cyclisch patroon, om de stabiliteit van het lopen bij valide personen te verstoren (zie figuur 2.1). In reactie op deze verstoring gingen proefpersonen lopen met kortere, snellere en bredere stappen, waarbij de loopsnelheid constant bleef. Daarnaast
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werden de proefpersonen lokaal minder stabiel (λs werd groter) en vergrootten zij hun ML en BW MoS. Volgens de onderliggende theoretische modellen kan deze toename van de ML en BW MoS verklaard worden als een gevolg van de gevonden aanpassingen in de stapparameters. Op basis van de resultaten van deze studie was het niet mogelijk om te concluderen in hoeverre de toename in λs puur een gevolg was van de verstoring, of in hoeverre de aanpassingen van het looppatroon deze stijging gedeeltelijk teniet heeft gedaan.

**Resons op de adaptatietaak**

In hoofdstuk 3 is een adaptatietaak gebruikt om te achterhalen welke strategieën valide personen gebruiken om de balans te handhaven tijdens het lopen. Tijdens het uitvoeren van deze adaptatietaak waren snelle en accurate adaptaties van het looppatroon noodzakelijk om virtuele targets die op het scherm verschenen te raken (zie figuur 3.1). Om een potentieel conflict tussen de regulatie van de stabiliteit van het lopen en de regulatie van het adaptatievermogen tijdens het lopen vast te stellen, werd de adaptatietaak zowel met als zonder de verstoring uit hoofdstuk 2 aangeboden. Tijdens het uitvoeren van de adaptatietaak, zowel met als zonder verstoring, gingen proefpersonen langzamer lopen als gevolg van een kleinere staplengte bij gelijkblijvende stapfrequentie. Daarnaast gingen zij enigszins breder lopen. Opnieuw kunnen deze aanpassingen in het looppatroon verklaard worden als mogelijke strategie om de ML en BW MoS te behouden tijdens het lopen. Wanneer we de aanpassingen van het looppatroon in reactie op de adaptatietaak (hoofdstuk 3) vergelijken met de aanpassingen in reactie op de translatoire platformverstoring (hoofdstuk 2), valt het op dat in reactie op de adaptatietaak een toename in stapfrequentie achterwege blijft. Een mogelijke verklaring hiervoor is dat een toename in stapfrequentie de tijd om te reageren op de virtuele targets verkort, wat mogelijk een nadelig effect heeft op de accuraatheid van de uitvoering van de adaptatietaak.

**Respons op manipulaties van het looppatroon**

Om te onderzoeken in hoeverre de gevonden aanpassingen van het looppatroon in reactie op de platformverstoring en de adaptatietaak een directe oorzaak zijn van de veranderingen in λs en de ML en BW MoS, hebben we in hoofdstuk 4 stapfrequentie, staplengte en loopsnelheid systematisch gemanipuleerd. De lokaal dynamische stabiliteit, uitgedrukt in λs veranderde niet als gevolg van de opgelegde manipulaties. De ML MoS namen toe bij een toename in stapfrequentie, terwijl de BW MoS toenam bij een afname in loopsnelheid en bij een afname in staplengte. Deze resultaten waren in lijn met onze verwachtingen gebaseerd op de zogenaamde ‘inverted pendulum’ modellen. Ondanks het feit dat deze modellen het
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lopen sterk simplificeren, blijken deze modellen geschikt om de MoS tijdens het lopen te voorspellen.

Conclusies
De aanpassingen in het looppatroon in respons op de platformverstoring hebben het doel om de ML and BW MoS te vergroten, wellicht om te compenseren voor de afname in de lokaal dynamische stabiliteit als gevolg van de verstoring. Een toename in stapfrequentie en stapbreedte, een afname in staplengte en het gelijk blijven van loopsnelheid hebben als direct gevolg dat zowel de ML als de BW MoS toenemen. In reactie op de adaptatietaak, zowel met als zonder platformverstoring, waren de valide proefpersonen in staat om de snelle en accurate adaptaties uit te voeren terwijl zowel de ML als de BW MoS werden behouden door aanpassingen in het looppatroon.

De resultaten van de studies met valide proefpersonen bieden een referentie wat betreft de strategieën die gebruikt kunnen worden in uitdagende loopsituaties om de stabiliteit te behouden of mogelijk te vergroten. In hoeverre personen met een beperking van de loopvaardigheid in staat zijn om dezelfde strategieën te gebruiken als valide personen om verstoringen van stabiliteit en manipulaties die een beroep doen op het adaptatievermogen te weerstaan, en in hoeverre verschillen in het looppatroon tussen deze groepen tijdens onverstoorde lopen kunnen worden toegeschreven aan de regulatie van de MoS is onderzocht in de daaropvolgende studies.

Stabiliteit en adaptatievermogen tijdens het lopen bij personen met een beperking van de loopvaardigheid
In hoofdstuk 5 en 6 hebben we een antwoord gegeven op de derde onderzoeksvraag van dit proefschrift:

3. Welke strategieën gebruiken personen met een onderbeen amputatie en personen na een CVA om manipulaties van stabiliteit en het adaptatievermogen van het lopen te weerstaan, en in hoeverre verschillen deze strategieën van de strategieën die worden gebruikt door valide personen?

In deze studies hebben we eerst onderzocht in hoeverre er verschillen bestaan in het looppatroon tijdens het onverstoorde lopen tussen personen met een onderbeenamputatie en personen na een CVA aan de ene kant en valide personen aan de andere kant. Vervolgens hebben we de respons van beide patiëntengroepen op de stabiliteitsmanipulatie en de adaptatietaak vergeleken met de respons van de valide personen. Door het vergelijken van de verschillen in deze respons met de verschillen die we hebben gevonden tijdens het
onverstoorde lopen tussen patiënten en valide personen, is het mogelijk om te achterhalen in hoeverre deze verschillen in het looppatroon gezien kunnen worden als een strategie om stabiliteit van het lopen te handhaven.

**Verschillen in het onverstoord lopen**

Zowel de personen met een onderbeenamputatie als personen na een CVA liepen langzamer dan valide personen. Voor de amputatiegroep was dit vooral te wijten aan een lagere stapfrequentie. Staplengte verschilde niet tussen de amputatiegroep en de groep valide personen. De lagere loopsnelheid resulteerde daarom ook in een kleinere BW MoS voor de amputatiegroep, wat in overeenstemming is met de resultaten beschreven in hoofdstuk 4. In tegenstelling tot de amputatiegroep, liepen de proefpersonen na CVA met name langzamer door een kortere staplengte. Omdat een kortere staplengte een positief effect heeft op de BW MoS, veroorzaakte de lagere loopsnelheid enkel een tendens in de richting van een kleinere BW MoS voor de CVA-groep in vergelijking met de valide groep.

Zowel de personen met een onderbeenamputatie als de personen na CVA liepen met een grotere stapbreedte in vergelijking tot de valide groep. Voor de amputatiegroep resulteerde dit in een grotere ML MoS dan bij de valide personen, ondanks de lagere stapfrequentie. Deze grotere ML MoS is mogelijk een compensatie voor de lagere lokaal dynamische stabiliteit (de hogere $\lambda_s$) die gevonden is in de amputatiegroep. Voor de personen na CVA was er geen verschil in de ML MoS vergeleken met de valide personen, ook al liepen deze personen met een grote stapbreedte en een relatief hoge stapfrequentie. De reden dat deze aanpassingen in het looppatroon niet hebben geleid tot een toename in ML MoS was de grote zijwaartse beweging van het bekken bij de personen na CVA, wat resulteerde in een grotere amplitude van het XCoM.

**Aanpassingen van het looppatroon als respons op de balansverstoring en de adaptatietak**

De aanpassingen van het looppatroon in reactie op de balansverstoring en de adaptatietak verschillen niet tussen de personen met een onderbeenamputatie en de valide personen. Daarnaast waren deze aanpassingen in overeenstemming met de aanpassingen van valide personen gevonden in hoofdstuk 2 en 3. Voor beide groepen resulteerden de aanpassingen in het looppatroon in een toename van de ML en BW MoS in reactie op de platformverstoring en het behoud van de ML en BW MoS in reactie op de adaptatietak. Ook de nauwkeurigheid waarmee de targets werden geraakt met de knieën verschilde niet tussen beide groepen.

In tegenstelling tot de personen met een onderbeenamputatie, verschilde de respons van de personen na CVA op de manipulaties wel van de respons van de valide personen. Personen na CVA verlaagden hun loopsnelheid, die al laag was tijdens het onverstoorde lopen, nog
verder in reactie op beide manipulaties. Deze afname in loopsnelheid zou een negatief effect kunnen hebben op de BW MoS. Echter, tijdens de platformverstoring ging de afname in loopsnelheid gepaard met een relatief grote afname in staplengte, waardoor een afname in BW MoS als gevolg van deze verstoring achterwege bleef. Tijdens het uitvoeren van de adaptatietaak, was deze afname in staplengte minder groot, waardoor de BW MoS tijdens deze taak kleiner was voor de CVA-groep in vergelijking met de valide personen. Daarnaast was ook de nauwkeurigheid waarmee de adaptatietaak werd uitgevoerd slechter in de CVA-groep dan in de valide groep.

**Conclusies**

Op basis van de resultaten gepresenteerd in hoofdstuk 5 en 6 kunnen we concluderen dat de personen met een onderbeenamputatie die hebben deelgenomen aan de studie, de mogelijkheid hebben om, net als valide personen, een strategie te selecteren die de ML en de BW MoS vergroot tijdens balansverstoringen, en de ML en BW MoS behouden tijdens het uitvoeren van een taak waar snelle en accurate aanpassingen van het looppatroon noodzakelijk zijn. Omdat de proefpersonen met een amputatie in staat waren om onder andere de BW MoS te vergroten tijdens het lopen met de balansverstoring, lijkt de kleinere BW MoS voor deze groep tijdens onverstoord lopen geen gevolg van een gebrek aan regelmogelijkheden van deze BW MoS. Mogelijk is deze kleinere BW MoS een neveneffect van andere beperkingen van het lopen met een prothese. Wellicht selecteren mensen met een onderbeenprothese een lagere loopsnelheid ter compensatie van de hogere energetische kosten van het lopen met een prothese, ook al gaat dit ten koste van de grootte van de BW MoS.

In tegenstelling tot de personen met een onderbeenamputatie hadden de personen na CVA wel moeilijkheden met de regulatie van de BW MoS, vooral tijdens het uitvoeren van de adaptatietaak. Daarnaast was de nauwkeurigheid waarmee CVA-groep de adaptatietaak uitvoerde slechter in vergelijking met de valide groep. Dit kan er op wijzen dat in het dagelijks leven, waarbij aanpassingen van het looppatroon bijvoorbeeld noodzakelijk zijn om obstakels te ontwijken, personen na een CVA een groter risico op vallen hebben door zowel een grotere kans op het raken van de obstakels als een grotere kans op het verliezen van de balans in achterwaartse richting. De resultaten van deze studie zijn een aanwijzing dat het trainen van het aanpassen van stapfrequentie en lengte terwijl loopsnelheid constant wordt gehouden bij personen na een CVA mogelijk een positief effect heeft op de capaciteit van deze personen om BW MoS te reguleren tijdens het lopen. In dit geval kunnen dergelijke trainingen personen na een CVA helpen om een looppatroon te selecteren dat hen robuuster
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maakt tegen verstoring die de potentie hebben om een val achterover te veroorzaken, zoals het uitlijden door een gladde ondergrond.

De rol van een asymmetrie in staplengte in de regulatie van de ‘backward margins of stability’ bij personen met onderbeenamputatie

In de voorafgaande paragrafen van deze samenvatting werd met betrekking tot de aanpassingen van het looppatroon geen onderscheid gemaakt tussen het aangedane en het niet-aangedane been. Echter, één van de karakteristieken van bijvoorbeeld personen met een onderbeenamputatie, is een asymmetrisch looppatroon. Zo is de stap met het niet-aangedane been (waarbij het niet-aangedane been leidend was en dus met het prothesebeen wordt afgezet) vaak korter in vergelijking met de andere stap. Een dergelijke asymmetrie wordt vaak gezien als een nadelig gevolg van het lopen met een prothese, maar een kortere stap met het niet-aangedane been kan mogelijk ook voordelen hebben met betrekking tot de BW MoS. Op basis van de resultaten gepresenteerd in hoofdstuk 7 hebben we een antwoord geformuleerd op de vierde onderzoeksvraag van dit proefschrift:

4. Op welke manier draagt de asymmetrie in staplengte bij personen met een onderbeenamputatie bij aan de regulatie van de BW MoS?

Uit de resultaten van deze observationele studie, bleek dat een kortere stap met het niet-aangedane been en nog specifieker een kortere voorwaartse voetplaatsing (de afstand tussen de voorste voet en het lichaamszwaartepunt bij hielcontact), resulteerde in een grotere BW MoS op hielcontact in vergelijking met de stap waarin het prothesebeen leidend is. Daarnaast vonden we slechts een beperkte toename in voorwaartse snelheid van het lichaamszwaartepunt tijdens de daaropvolgende transitie van de niet-aangedane stap naar de prothese stap. De grotere BW MoS op hielcontact van de stap van het niet-aangedane been is daarom mogelijk een compensatie om er voor te zorgen dat de BW MoS ook aan het einde van deze transitie nog groot genoeg is om een verlies van balans in achterwaartse richting te voorkomen en de voorwaartse looptoerusting te handhaven. Op basis van de resultaten van deze studie is het niet mogelijk om te concluderen of en in hoeverre de regulatie van de BW MoS de primaire reden is van de asymmetrie in staplengte, maar deze observatie maakt in ieder geval duidelijk dat een asymmetrie niet enkel een nadelig gevolg van het lopen met een prothese hoeft te zijn. Het nastreven van een symmetrisch looppatroon tijdens de revalidatie zou daarom niet per sé een doel moeten zijn en heeft misschien zelfs nadelige gevolgen.
Algemene conclusies

Het onderzoek gepresenteerd in dit proefschrift draagt bij aan een beter begrip van de manier waarop aanpassingen in het looppatroon kunnen bijdragen aan het behoud of zelfs het verbeteren van de stabiliteit tijdens het lopen in uitdagende omstandigheden. We hebben aangetoond dat sommige verschillen in het onverstoorde lopen tussen personen met een beperking van de loopvaardigheid en valide personen, zoals een grotere stapbreedte of een asymmetrie in staplengte bij personen met een amputatie, functioneel kunnen zijn voor de regulatie van de MoS. Andere verschillen in het looppatroon, zoals een lagere loopsnelheid bij zowel personen met een amputatie als personen na CVA lijken op dit vlak juist niet functioneel te zijn.

Met name personen na CVA blijken moeite te hebben met het selecteren van een strategie die resulteert in het behoud van met name de BW MoS, onder uitdagende loopomstandigheden. Vervolgonderzoek zal moeten uitwijzen waarom personen na CVA een dergelijk strategie niet gebruiken. Daarnaast kan het van belang zijn om te onderzoeken in hoeverre specifieke trainingen gericht op het aanpassen van stapfrequentie en staplengte kunnen helpen om de stabilititeit tijdens het lopen in het dagelijks leven te behouden om op deze manier de kans op vallen te reduceren.
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ABOUT THE AUTHOR
Curriculum Vitae

Laura Hak was born on March 4, 1981 in Alkmaar (The Netherlands). After finishing high school in 1999 (Jan Arentsz college, Alkmaar), she started studying Human Movement Sciences at the VU University in Amsterdam. During her study she works as a student assistant in the Biomechanics course. In 2005, she graduated with an accent in Sports Psychology and a teaching qualification. From 2004 until 2010 she developed and coordinated a digital pre-master curriculum of the Faculty of Human Movement Sciences. Besides, from 2007 until 2010 she worked as a teacher in Biomechanics, Physiology and Statistics at the Academy of Physical Education in Amsterdam. In February 2010 she started with her Ph.D. project on the Biomechanics of walking, and more specific on the risk of falling during walking, in amputees and post-stroke individuals. This project resulted in six peer reviewed publications, so far. The project was a cooperation between the MOVE Research Institute Amsterdam, Motek Medical B.V. Amsterdam, and the Military Rehabilitation Center in Doorn. During this project she presented at multiple international congresses like the congress of the International Society of Biomechanics (Brussels, 2011), the congress of the International Society for Posture & Gait Research (Trondheim, 2012), and the congress of the European Society for Movement Analysis in Adults and Children (Glasgow, 2013). In the immediate future Laura aims to continue her activities as a researcher and a teacher in the field of Biomechanics an Rehabilitation. The coming year she will continue working at the Faculty of Human Movement Sciences (VU University Amsterdam) combining research and teaching.
List of publications

*Peer reviewed international journals*


**Hak L, Van Dieën JH, Van der Wurff P, Houdijk H.** Step length asymmetry in transtibial amputees; a strategy to regulate the margins of stability? Phys Ther (Accepted for publication).

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National conference abstracts
