



Stepping strategies for regulating gait adaptability and stability



Laura Hak^a, Han Houdijk^{a,b}, Frans Steenbrink^c, Agali Mert^d, Peter van der Wurff^{d,e},
Peter J. Beek^a, Jaap H. van Dieën^{a,*}

^a Research Institute MOVE, Faculty of Human Movement Sciences, VU University, Van der Boechorststraat 9, 1081 BT Amsterdam, The Netherlands

^b Helimare Rehabilitation Centre, Wijk aan Zee, The Netherlands

^c Motek Medical b.v., Amsterdam, The Netherlands

^d Center for Augmented Motor Learning and Training, National Military Rehabilitation Centre Aardenburg, Doorn, The Netherlands

^e Pain Centre Department of Anaesthesiology, University Medical Centre Groningen, University of Groningen, The Netherlands

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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.

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1. Introduction

For patients with gait impairments, falling during walking is one of the main problems (Miller et al., 2001a, 2001b). Knowledge of strategies that patients could use to minimise 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 (Hak et al., 2012; McAndrew et al., 2010a, 2010b). 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 (Hak et al., 2012; McAndrew et al., 2010a, 2010b). As changing walking speed has been proposed to be an important strategy to enhance gait stability (Dingwell and Marin, 2006; England and Granata, 2007; Stergiou et al., 2004; Weerdesteyn et al., 2008), we previously (Hak et al., 2012) 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

* Corresponding author.

E-mail address: j.van.dieen@vu.nl (J.H. van Dieën).

backward (BW) direction, and therefore decrease the risk of falls in these directions (Hof et al., 2005, 2007). 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 (Hak et al., 2012).

Studies that used protocols to provoke an adaptive gait pattern mostly used an obstacle avoidance task (Den Otter et al., 2005; Hofstad et al., 2006; Said et al., 2008, 1999). In some of these studies (Den Otter et al., 2005; Hofstad et al., 2006) 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 destabilising effect of this movement (Chou et al., 2001; Said et al., 2008, 1999).

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 hypothesised 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 hypothesised that subject would decrease step length and increase step width in response to the GA-task (Hak et al., 2012; McAndrew et al., 2010b). An increase in step frequency, previously found as strategy to enhance stability (Hak et al., 2012; McAndrew et al., 2010b), 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 (Bishop et al., 2004; Chen et al., 1994; Fitts, 1954). Therefore, we hypothesised 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 (Hak et al., 2012). We hypothesised 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 destabilising effect of the perturbation, while the expected lack of an increase in step frequency during the GA-task would hamper balance control.

2. Methods

2.1. 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.

2.2. 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) (Fig. 1a). VE's were projected onto a 180° 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 (Fig. 1b). The motion platform was used to induce perturbations in ML-direction in selected

trials (Fig. 1c). D-flow software was used to control the system and to synchronise the instantaneous treadmill speed and scene progression (Geijtenbeek et al., 2011). 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 antero-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.

2.3. Protocol

2.3.1. 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.

2.3.2. Experimental trials

The protocol consisted of four trials of 4 min 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 a 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:

$$D(t) = 0.05[1.0\sin(0.16 \times 2\pi t) + 0.8\sin(0.21 \times 2\pi t) + 1.4\sin(0.24 \times 2\pi t) + 0.5\sin(0.49 \times 2\pi t)] \quad (1)$$

where $D(t)$ is the translation distance (m) and t is time (s) (Hak et al., 2012; McAndrew et al., 2010a, 2010b).

2.4. 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 epicondylus 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.

2.5. Data analysis

2.5.1. Walking speed

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

2.5.2. 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.

2.5.3. Margins of stability

To calculate the margins of stability (MoS), a method derived from the procedure developed by Hof et al. (2005, 2007) 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

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