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

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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 an overall smaller backward margins of stability, and a significantly smaller backward margin 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 margin of stability when walking in complex environments.

1. Introduction

People who are recovering from a stroke have an increased risk of falling during walking (Weerdesteyn et al., 2008). In the literature several causes for this increased risk of falling are suggested, such as an enlarged body sway in the fontal plane during steady state walking (De Bujunda et al., 2004; Tyson, 1999), and a limited capacity to adapt the gait pattern in response to environmental demands, for example to avoid an obstacle (Den Otter et al., 2005; Said et al., 1999, 2008). 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 (Den Otter et al., 2005). 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 (Said et al., 2008).

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 (Hof et al., 2005). The MoS can be calculated in both medio-lateral (ML) (Hof et al., 2007; McAndrew Young & Dingwell, 2012; McAndrew Young et al., 2012) and anterio-posterior (AP) (Espy et al., 2010b; McAndrew Young & Dingwell, 2012; McAndrew Young et al., 2012; Pai & Patton, 1997 Apr) 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 (Bierbaum et al., 2010, 2011; Hak et al., 2012; McAndrew Young et al., 2012). MoS can be regulated effectively by adjusting step parameters. From studies of Hof

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et al. (Hof, 2008; Hof et al., 2005, 2007), 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. (Espy et al., 2010a, 2010b) 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 not only 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 (Hak et al., 2013b; Hak et al., 2012; McAndrew Young et al., 2012). 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 (Hak et al., 2013a,b). 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 (Hak et al., 2013a). 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 (Hak et al., 2013a,b).

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 (Chen et al., 2005; Krasovsky et al., 2012). In the study of Chen et al. (Chen et al., 2005), 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 (Dingwell & Marin, 2006; England & Granata, 2007; Kang & Dingwell, 2008; Krasovsky et al., 2012). However, a lower walking speed may decrease the BW MoS (Espy et al., 2010a; Pai & Patton, 1997). 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 (Hof et al., 2005; Hof et al., 2007). 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 (Hak et al., 2012; Hak et al., 2013a,b). 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.

2. Methods

2.1. Subjects

Ten adult subjects who had suffered from a stroke (age mean 60.8 (SD 8.4) years, height mean 1.79 (SD 0.07) m, mass mean 88.4 (SD 8.5) kg) and 9 age-matched control subjects (age mean 57.3 (SD 7.2) years, height mean 1.77 (SD 0.08) m, mass mean 79.7 (SD 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 Centre Aardenburg, Doorn, The Netherlands. A minimum score of 4 on the Functional Ambulation Categories (FAC) (Holden et al., 1986) in combination with a minimum score of 45 on the Berg Balance Scale (BBS) (Stevenson, 2001) was required to participate in this study. Further characteristics of the post-stroke group are reported in Table 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.

2.2. 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) (Fig. 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 (Davis et al., 1991; Kadaba et al., 1990)). 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.

2.3. Protocol

2.3.1. Familiarization

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

2.4. Experimental trials

The actual protocol consisted of 3 trials of 4 minute 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 (Hak et al., 2012; McAndrew et al., 2010a, 2010b) (Fig. 1B).

For the gait adaptability task (GA task) the VE was used to project targets on the screen (Fig. 1C). 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 virtual obstacle avoidance tasks is the