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More symmetrical gait after split-belt treadmill walking does not modify dynamic and postural balance in individuals post-stroke

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ABSTRACT

Spontaneous gait is often asymmetrical in individuals post-stroke, despite their ability to walk more symmetrically on demand. Given the sensorimotor deficits in the paretic limb, this asymmetrical gait may facilitate balance maintenance. We used a split-belt walking protocol to alter gait asymmetry and determine the effects on dynamic and postural balance. Twenty individuals post-stroke walked on a split-belt treadmill. In two separate periods, the effects of walking with the non-paretic leg, and then the paretic one, on the faster belt on spatiotemporal symmetry and balance were compared before and after these perturbation periods. Kinematic and kinetic data were collected using a motion analysis system and an instrumented treadmill to determine symmetry ratios of spatiotemporal parameters and dynamic and postural balance. Balance, quantified by the concepts of stabilizing and destabilizing forces, was compared before and after split-belt walking for subgroups of participants who improved and worsened their symmetry. The side on the slow belt during split-belt walking, but not the changes in asymmetry, affected balance. Difficulty in maintaining balance was higher during stance phase of the leg that was on the slow belt and lower on the contralateral side after split-belt walking, mostly because the center of pressure was closer (higher difficulty) or further (lower difficulty) from the limit of the base of support, respectively. Changes in spatiotemporal parameters may be sought without additional alteration of balance during gait post-stroke.

1. Introduction

Individuals post-stroke have activity limitations related to various locomotor impairments, such as reduced walking speed (Balaban and Tok, 2014; Richards et al., 2015), asymmetrical gait pattern (Balaban and Tok, 2014; Patterson et al., 2008), and static (Tasseel-Ponche et al., 2015) and dynamic balance deficits (Kao et al., 2014; Nott et al., 2014). Post-stroke gait is less stable during the paretic stance phase as revealed by the alteration of the displacements of the centre of pressure under the paretic foot (Chisholm et al., 2011) or the increased angular momentum in the frontal plane during paretic stance phase (Nott et al., 2014). Although variable among individuals post-stroke, balance deficits could explain in part their reduced gait speed and increased fall risk (Weerdesteyn et al., 2008).

Asymmetry of spatiotemporal (ST) gait parameters (e.g. step length, stance time and swing time) has also been shown to be related to reduced gait speed (Patterson et al., 2010), decreased standing balance (Hendrickson et al., 2014) and clinical scores of balance assessment (Lewek et al., 2014). This raises the question whether asymmetrical gait may facilitate balance compared to a more symmetrical gait, in part because of the sensorimotor deficits at the paretic lower-limb.

Asymmetry in step length (SL) and double support time (DST) can be reduced by using a split-belt treadmill protocol in individuals post-stroke (Lauziere et al., 2014; Reisman et al., 2007). Split-belt walking requires a reorganization of the locomotor pattern shown by the alteration of most ST parameters. Following several minutes of perturbation at unequal speeds, changes in SL and DST are maintained for some cycles once the belts return to equal speeds, contrary to other parameters, such as stance time, that immediately come back to pre-perturbation values. This protocol improved SL symmetry when the faster belt was used on the side with shorter SL during split-belt walking after stroke. It also changed the generation of moments at the ankle, with higher plantarflexion moments on the slow belt, and lower moments on the fast belt in post-perturbation (Lauziere et al., 2014). Currently, we do not know whether the changes in SL symmetry and joint moments observed after split-belt walking affect balance. Since these changes influence foot placement and base of support (BoS) configuration in anteroposterior direction (Balasubramanian et al.,

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2010; Hak et al., 2013), one can expect an impact on dynamic balance, because balance control is determined by foot placement and the position of the center of mass (CoM) relative to the feet (Winter, 1995). Moreover, foot placement allows the control of the position of the center of pressure (CoP) and the CoP position relative to the CoM controls the CoM accelerations, i.e. the further apart the CoP and the CoM, the larger the acceleration of the CoM (Winter, 1995). Thus, alteration of the symmetry of the ST parameters likely affects balance during gait in individuals post-stroke. Understanding how changes in the symmetry of gait affect balance is important to better support the use of split-belt training, and more generally of interventions to improve symmetry in walking post-stroke.

The first objective of the study was therefore to determine the effect of changes in ST symmetry, induced after split-belt walking, on the difficulty in maintaining balance in individuals post-stroke. It was assumed that dynamic and postural balance would be more difficult to maintain when these individuals walked more symmetrically, which could be one factor explaining why an asymmetrical gait pattern is spontaneously used instead of a more symmetrical one. Considering that balance impairment was observed more during the paretic than non-paretic stance phase, and because the effects of split belt on moment generation and asymmetry differ according to which leg is placed on the faster belt during perturbation (Lauziere et al., 2014), a second objective was to identify which of the split-belt treadmill conditions, i.e. the non-paretic or paretic leg on the fast belt, most affects dynamic and postural balance.

2. Methods

2.1. Participants

Twenty participants (mean age: 49.4 years, standard deviation (SD): 13.2; 13 men) who had their first unilateral supratentorial stroke (14 right-side lesion) more than 6 months ago (mean: 84.4 months, SD: 93.1) were recruited in this study. They were included if they were able to walk independently 10 m overground at a gait speed ≥ 0.5 m/s without assistive devices or physical assistance. They were excluded if they had a cerebellum lesion or any cognitive or medical conditions that could affect their locomotor ability. All participants signed a consent form approved by the local ethics committee.

2.2. Clinical assessment

Participants' self-selected and maximal overground gait speed, functional mobility, balance, and leg/foot motor recovery were evaluated using the 10-meter walk test (Salbach et al., 2001), the Timed Up and Go test (Ng and Hui-Chan, 2005), the Berg Balance Scale (Berg et al., 1995) and the Chedoke McMaster Stroke Assessment (Gowland et al., 1993), respectively. To determine self-selected speed on the treadmill, the participants walked with the speed increasing by 0.1 m/s every 45 s, until the speed was deemed uncomfortable by the participant. The prior speed was considered the treadmill self-selected speed for this participant.

2.3. Experimental protocol

The participants walked on the split-belt treadmill following a previously used protocol (Lauziere et al., 2014): (1) baseline, tied belt at self-selected speed, for 3 min; (2) perturbation, split belt with the slow belt at self-selected speed and the fast belt at double the self-selected speed, for 6 min; and (3) post-perturbation, tied belt at self-selected speed for 3 min (idem as baseline). They experienced this protocol twice, first with the non-paretic leg on the fast belt (NP Fast condition), then with the paretic leg on the fast belt (P Fast condition) with 10 min of rest between protocols. During all periods, for safety reasons, participants wore a harness that did not provide weight support. Participants held side-mounted handrails only during the perturbation period. Fifteen consecutive gait cycles were analyzed from the baseline period and at the beginning of the post-perturbation period, i.e. immediately after the perturbation period ended and the participant released the handrail (in the first cycle post-perturbation).

2.4. Data collection

A 3D whole-body motion analysis system, i.e. four Optotrak Certus cameras (Northern Digital Inc., Waterloo, ON, Canada) and three to six infrared markers placed on each body segment, was used to estimate body CoM kinematics. The contour of the BoS was determined by the foot contours digitized relative to the foot markers. The instrumented split-belt treadmill (Bertec Fully Instrumented Treadmill (FIT®)) recorded ground reaction forces and the global CoP at a frequency of 600 Hz and these signals were re-sampled at 60 Hz to match the kinematic data. Kinematic and kinetic data were used to quantify difficulty in maintaining balance by using the concepts of stabilizing and destabilizing forces (Duclos et al., 2009; Duclos et al., 2012).

The stabilizing force represents the dynamic component, i.e. the theoretical force needed to stop the body movement (CoM velocity) at the limit of the BoS, while the destabilizing force represents the postural component, i.e. the theoretical force needed to bring the CoP to the limit of the BoS. Higher stabilizing force (i.e. higher CoM velocity or shorter CoP-BoS distance) and lower destabilizing force (i.e. lower weight, higher CoM or shorter CoP-BoS distance) indicate greater difficulty in maintaining balance during the task (Duclos et al., 2009; Duclos et al., 2012). This model is sensitive to reduced proprioceptive integration in balance control during walking in individuals post-stroke (Mullie and Duclos, 2014) and changes in difficulty level of balance perturbations in healthy participants (Ilmane et al., 2015). It was also used to show that individuals with spinal cord injury reduced their walking speed to ensure their balance when walking overground (Lemay et al., 2014) or on an inclined pathway (Desrosiers et al., 2014). One variable included in the calculation of both forces is the distance between the CoP and the BoS (CoP-BoS distance) in the direction of CoM velocity (Duclos et al., 2012). It represents the distance available to generate a postural reaction and is not included in other tools to evaluate balance such as the extrapolated center of mass (Hof et al., 2005). Another variable included in the stabilizing force is the CoM velocity (Duclos et al., 2009). These variables (CoP-BoS distance and CoM velocity) were analyzed separately in addition to the stabilizing and destabilizing forces to further understand the determinants of balance difficulty. Peak values (maximum for stabilizing force, and minimum for destabilizing force) obtained during paretic and non-paretic stance phases of the gait cycle were used because they indicate the highest level of difficulty in managing balance during the stance phase (Duclos et al., 2009). Peak values were normalized to the body mass for the stabilizing and destabilizing forces.

Temporal (DST, swing and stance times) and spatial (SL, trunk progression and foot forward placement (Roerdink and Beek, 2011)) parameters were obtained using the vertical ground reaction forces and kinematics of the feet and pelvis. The paretic DST was defined as the time between non-paretic foot contact and subsequent paretic toe-off, and reciprocally for the non-paretic DST. The paretic swing time (SwT) was the time between the paretic toe-off and the subsequent paretic heel contact, while the paretic stance time (StT) was the time between the paretic foot contact and the subsequent paretic toe-off, and reciprocally for the non-paretic side. For the spatial parameters, the paretic/non-paretic SL was defined by the anteroposterior distance of the markers on the paretic and non-paretic lateral malleoli at paretic/non-paretic foot contact. Given the variation of the asymmetry of the SL, the model suggested by Roerdink and Beek (2011) for measuring the SL symmetry taking into account the trunk progression (TP) and