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Young and old adults prioritize dynamic stability control following gait perturbations when performing a concurrent cognitive task

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ABSTRACT

There are conflicting results regarding the effect of aging on postural prioritization. The present study investigated balance recovery performance of young and old adults following unexpected gait perturbations in a dual task condition. Thirty-two young and 30 elderly male subjects were assigned to either control or dual task group. After baseline assessment, an unexpected gait perturbation was induced by a sudden change of surface rigidity. The dual task groups performed a mental arithmetic task. Dynamic stability was quantified based on the 'extrapolated center of mass' concept. The margin of stability decreased significantly at touchdown of the recovery leg following the unexpected perturbation (P) compared with baseline (base), yet irrespective of cognitive load (base: −4.63 cm; P: −13.32 cm; $p < 0.05$). The number of errors in the cognitive task increased significantly (base: 0.13; P: 0.48; $p < 0.05$) in both age groups. Since the stability performance was unaffected by additional cognitive load, whereas the cognitive task performance declined following the perturbation in both groups, it is concluded that postural prioritization occurs independent of age in response to unexpected gait perturbations.

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1. Introduction

The age-related decrease of balance recovery abilities, associated with an increase of fall incidents with age [1], is commonly attributed to degenerative processes of the neuromuscular system [2–4]. However, results of Arampatzis et al. [5] suggest that, within an elderly subject population, the deficits in the recovery from a forward fall may not relate solely to muscle-tendon unit capacities but the selection and application of appropriate motor programs.

A growing body of evidence suggests, that high-level processing is involved in the modulation of postural responses (reviewed in Ref. [6]). However, processing resources are limited and since everyday activity is characterized by multitasking, this may induce interference effects due to central overload [7,8]. Task prioritization then affects the direction of the interference [9]. With regard to the decline of processing resources with age [10], it seems promising to focus research on the effects of a dual task related increase of processing requirements on locomotor stability control in the elderly. By some authors it is referred to as "posture first strategy", when cognitive task performance is compromised to secure postural stability [11], yet inconsistent results were obtained with regard to the effect of age on task prioritization [12–15]. However, to the best of our knowledge there is no study where locomotion was as severely perturbed as it may precede fall incidents while taking into account the effect of dual tasking. Therefore, the objective of the current study is to examine the effect of cognitive load on dynamic stability responses to unexpected gait perturbations in old and young adults. We hypothesize that, when subjected to a severe postural threat, (a) a concurrent cognitive task would not affect dynamic stability and (b) this behavior would not be affected by age.

2. Method

2.1. Subjects

Thirty-two young and 30 elderly male, healthy and physically active subjects participated in the present study and were randomly assigned to control (young: $n = 15$, mean ± SD values of age 26.2 ± 3.2 years, body height 183.5 ± 6.7 cm, weight 75.3 ± 8.3 kg; old: $n = 16$, 69.4 ± 4.3 years, 177.4 ± 7.4 cm, 80.2 ± 8.6 kg) or dual-task group (young: $n = 17$, 27.9 ± 1.8 years, 180.6 ± 7.5 cm, weight 75.8 ± 8.7 kg; old: $n = 14$, 68.6 ± 3.3 years, 173.6 ± 6.0 cm, 77.9 ± 9.0 kg). All subjects signed a written approval consent form concerning the procedure, which was also approved by the Internal Ethical Committee.

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2.2. Experimental design

Beneath a covered exchangeable element (0.6 m × 0.8 m), positioned halfway along a gangway (15 m × 0.6 m × 0.2 m), a foam block featuring a non-linear force-deformation-characteristic provided the compliant surface for the perturbation trial. The subjects were secured by means of a body harness.

The target walking velocity (self-selected brisk velocity), monitored by three light barriers enclosing the first half of the gangway, was determined individually for every participant. After some practice trials the average velocity of three trials was defined as target velocity. The starting position was adjusted so the subject would step on the removable element with the right leg without altering spatiotemporal gait characteristics.

Between each trial of the experimental protocol, the subjects rested inside an enclosed cabin. All participants performed three baseline trials with target velocity. The dual task group performed another three steady-state gait trials with the concurrent cognitive task (baseline-DT). Subsequently, the unexpected gait perturbation was induced in all groups (i.e. with concurrent cognitive task in the dual task group).

In the cognitive task, a starting value and a randomized sequence of seven auditory cues, including three different signals each assigned to a respective mathematical operation, were presented to the subjects via a wireless headset. Signal 1, a telephone ringtone, was assigned to an addition and signal 2, a door bell, to a subtraction. Signal 3, a barking dog, was assigned to a plus/minus zero operation and was applied three times maximum, since an elevated count of this signal would have simplified the task. In an accustoming-phase, the level of difficulty for each subject was increased successively by changing the operands (+1/−1, +2/−1, +3/−2, +5/−3) until the highest level the subject could perform twice without fault was assessed.

2.3. Data acquisition and processing

For recording whole body kinematics, 13 reflective markers with a diameter of 14 mm were fixed at the following anatomical landmarks: left and right acromion, 7th cervical vertebra, joint line of elbow and wrist, greater trochanter, joint line of the knee, lateral malleolus and four markers defined the head segment. The position of two markers at the heel and the middle toe on each shoe were plotted in sketches of the shoes for calculating the base of support.

Kinematic data were recorded using a Vicon motion capture system (Version 1.4.1, Vicon Motion Systems, Oxford, UK) integrating eleven cameras (6× Vicon F20, 5× Vicon T20) operating at 250 Hz. The marker trajectories were smoothed using a bidirectional fourth order Butterworth low-pass filter routine with a cut-off frequency of 6 Hz. Segmental masses and the location of the segment centers of mass were calculated based on the data reported by Dempster et al. [16].

Dynamic stability was quantified in the sagittal plane based on the "extrapolated center of mass" concept proposed by Hof et al. [18], modeling the human body as an inverted pendulum. The margin of stability (b_x) describes the stability of a system as the distance between the anterior border (u_{max}) of the base of support and the position of the center of mass (CM) extrapolated in the direction of its velocity (extrapolated CM or X_{CM}, see Eq. (2)):

$$b_x = u_{max} - X_{CM} \quad (1)$$

The behavioral and mechanical characteristics that determine the stability of the system are thus reflected in the base of support (BoS) and the components that define the extrapolated CM position:

$$X_{CM} = \frac{P_{CM} + V_{CM}}{\sqrt{g/l}}, \quad (2)$$

Being the horizontal CM velocity (V_{CM}), the projection of the CM to the ground (P_{CM}), and the term $\sqrt{g/l}$ (i.e. eigen frequency of the system, g: acceleration of gravity, l: the distance between ankle joint and CM as length of the pendulum).

The margin of stability, BoS, X_{CM} and all its components were determined at touchdown of both the perturbed and the recovery leg. The anterior and posterior borders of the BoS were the anterior border of the toe of the leading leg at touchdown and the heel of the trail leg during steady-state single stance respectively. The extrapolated CM as well as the CM projection was calculated in relation to the posterior border of the BoS.

The cognitive task responses associated with the phase of potential gait disturbance and balance recovery were analyzed with respect to errors and response time.

2.4. Statistics

Statistical analysis was performed in SPSS (SPSS Inc., Version 17.0, Chicago, USA). The mean of three trials each was used as representatives for baseline as well as baseline-DT (i.e. baseline with concurrent cognitive task). The parameters of dynamic stability were analyzed by means of a three-way ANOVA with the fixed factors trial (baseline, perturbation), age (young, old) and condition (control, dual task). The horizontal CM velocity and body height were used as covariates. In case of significant interactions, the respective factors were examined separately in a two-way ANOVA. A two-way ANOVA (age by trial) was also applied for the analysis of response time; the number of errors was examined with a Wilcoxon signed-ranks and Mann–Whitney-*U*-test. The level of significance was set to $\alpha = 0.05$.

3. Results

There were no significant trial differences ($p > 0.05$) of the parameters of dynamic stability between baseline with (baseline-DT) and without concurrent task among the dual task groups. The baseline-DT values were selected as reference for the perturbation trial.

3.1. Touchdown of the perturbed leg

Comparing baseline and perturbation values, there was no trial effect ($p > 0.05$) on the stability parameters at touchdown of the perturbed leg, the instance prior to the perturbation. However, there was a significant age effect ($p < 0.05$) on the horizontal CM velocity ($F_{1,114} = 37.13$) and the term $\sqrt{g/l}$ ($F_{1,114} = 16.38$). The young subjects had a higher CM velocity and smaller term $\sqrt{g/l}$, indicating a longer pendulum length (Table 1). There was a significant age by condition interaction ($p < 0.05$) on the CM projection ($F_{1,114} = 5.16$), the extrapolated CM ($F_{1,114} = 4.7$) and the BoS ($F_{1,114} = 4.02$). In older adults, the position of the CM projection ($F_{1,54} = 6.39$) as well as the extrapolated CM ($F_{1,54} = 5.83$) was more anterior and the BoS values were higher ($F_{1,54} = 6.56$) in the control group compared to the dual task group. Additionally, in the control subjects the CM projection ($F_{1,56} = 8.36$) and the extrapolated CM ($F_{1,56} = 11.36$) were more anterior in the young compared with old controls (Table 1).

3.2. Touchdown of the recovery leg

There was a significant trial effect on the margin of stability and all components of dynamic stability ($p < 0.05$) at touchdown of the

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