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Research paper

Comparison of gait velocity and center of mass acceleration under conditions of disrupted somatosensory input from the feet during the navigation of obstacles in older adults with good and poor visual acuity

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ABSTRACT

Objectives: To investigate the effects of good binocular visual acuity (GBVA) versus poor binocular visual acuity (PBVA), gait velocity and center of mass (COM) acceleration were evaluated in elderly individuals under three conditions, including the navigation of obstacles while walking with altered sensory conditions in the feet.

Methods: Nineteen elderly Korean women from community housing were enrolled in this study; nine participants had a binocular visual acuity (BVA) that was equal to or less than 0.4 logarithm of the minimum angle of resolution (logMAR), and 10 participants had a BVA that was equal to or greater than 0.3 logMAR. Participants were fitted with an accelerometer over the L3 spinal process and then walked on a GAITRite® system at a self-selected speed under three different conditions (barefoot, wearing mountain socks, and stepping over obstacles while wearing mountain socks).

Results: The velocity of the GBVA group was significantly higher than was that of the PBVA group whereas the COM acceleration of the GBVA group was significantly lower than was that of the PBVA group. Both groups demonstrated significant differences in velocity and under the three experimental conditions and PBVA differed significantly under conditions 1 and 3 and under conditions 2 and 3.

Conclusions: These findings suggest that visual acuity (VA) in the elderly influences dynamic balance and gait velocity. Additionally, elderly participants with PBVA exhibited a greater sensitivity to altered sensory input, especially distorted sensory input from the feet in the presence of obstacles. Thus, elderly individuals with PBVA may require balance and gait training in diverse environments, including those involving the navigation of obstacles, to reduce the likelihood of falling.

1. Introduction

Balance is an essential component of the ability of elderly individuals to walk in a stable manner, negotiate obstacles, independently perform the activities of daily living (ADL), and maintain a good quality of life [1,2]. Balance is defined as the ability to maintain the projection of the body's centre of mass (COM) within manageable limits of the base of support (BOS). In addition, balance reflects the ability to appropriate static posture like standing or sitting, and dynamic stability to a new BOS during movement like walking [3].

The navigation of an obstacle is common challenge to the ability maintain balance during the performance of ADL and is a frequent cause of falls among older adults [4,5]. The navigation of obstacles necessitates greater joint motion in the swinging limb and greater joint kinetic demands in the standing limb compared with level walking. Enhanced motion in the swinging limb during obstacle crossing increases the need for balance and, as a result, BOS narrows and the COM move away from the BOS [4,6]. More complex and faster motion in the body segments while negotiating an obstacle will lead to greater and faster movement of the COM.

Afferent sensory input signals, such as from the visual, proprioceptive, and vestibular systems, play an important role when maintaining balance while navigating or stepping over obstacles [7,8]. Prior to crossing an obstacle, vision is required to detect the obstacle and evaluate its position, height, and size; this information is

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proactively used to plan a movement strategy [9]. Additionally, vision consistently provides updated information to the central nervous system (CNS) regarding the position and movement of body segments in relation to one another and the environment during the stabilization of postural control [10]. Somatosensory inputs from the body also contribute to postural control and bodily orientation. The sole of the foot is a particularly important contributor to proprioceptive input and provides information with respect to the contact surface in the service of maintaining postural control. Sensory information from the ankle also plays an important role during the adjustment of gait incidence and the avoidance of obstacles [11]. In turn, the CNS processes the afferent sensory information and enacts effective motor responses by adjusting the position of the lower extremities, instigating movement (such as range of motion, foot clearance, and foot placement), minimizing body sway, and regulating walking speed [5].

Age-related decreases in balance are influenced by inappropriate sensory input, and this requires that older adults adopt a conservative strategy related to gait and the navigation of obstacles [12]. Several variables have been used to quantify balance during walking and obstacle crossing [13–15], and the parameters of COM acceleration and walking speed have been shown to be sensitive measurements when quantifying movement pattern, balance problems, and falls in older adults [1,16,17]. Although healthy elderly adults adopt a more conservative strategy when negotiating obstacles in the environment by using slower crossing speed, a shorter step length, and a smaller step width compared with young adults, they still demonstrate an increased risk for obstacle contact. Hallemans et al. [18] also observed specific differences in the gait patterns of those with and without a visual impairment. These authors showed that, even in an uncluttered environment, vision is important for locomotive control and that visual impairments may result in a more cautious walking strategy that incorporates adaptive changes, such as using the foot to probe the ground for haptic exploration or having a greater dependence on tactile feedback from the plantar surface of the foot. These findings indicate that when a particular mode of sensory input is compromised, there is a greater reliance on other sensory cues to maintain balance. However, Lajoie et al. [19] evaluated the relative contributions of vision, proprioception, and other efferent inputs during the storage of a neural representation for guiding trail leg trajectory over an obstacle and found that altering proprioceptive feedback from the lead leg by adding mass to the ankle did not influence measures from the trail leg toe.

A number of studies have investigated the role of various sensory systems during the maintenance of static and dynamic balancing, but these have produced mixed results. Moreover, few studies have investigated the effects of gait velocity and COM during walking and the crossing over of obstacles under varied sensory conditions, including distorted sensory input from the feet. Therefore, the aim of this study was to evaluate gait velocity and COM in older adults with good or poor visual acuity (VA) during the navigation of an obstacle under conditions of distorted sensory input from the feet.

2. Materials and methods

2.1. Subjects

A power analysis was performed with G*power software ver. 3.1.2 (Franz Faul, University of Kiel, Kiel, Germany) using the results of a pilot study involving ten subjects. The calculation of sample size was carried out with a power of 0.80, alpha level of 0.05, and effect size of 1.91. This provided a necessary sample size of total twelve subjects for this study (GBVA = 6 subjects, PBVA = 6 subjects). A total of 19 elderly women from Gyeongsangnam-do, South Korea with varying levels of binocular visual acuity (BVA) participated in this study. The poor-BVA (PBVA) group consisted of nine participants with BVA that was lower than or equal to 0.4 logarithm of the minimum angle of resolution (logMAR), and the good-BVA (GBVA) group consisted of 10 individuals with BVA that was higher than or equal to 0.3 logMAR. The inclusion criteria were as follows: (a) older than 65 years of age with BVA of 0.4 or worse logMAR, (b) older than 65 years of age with BVA of 0.3 or better logMAR, (c) the ability to walk independently without an assistive device, and (d) a score >24 on the Korean Version of the Mini-Mental State Exam. The exclusion criteria were as follows: (a) a past or present neurological disorder such as stroke and Parkinson's disease (b) a major orthopedic diagnoses (bone fractures, joint fusions or replacements, limb amputations) in the lower back, pelvis and lower extremity, (c) significant auditory and/or vestibular impairments, (d) currently taking drugs (antidepressants, sedative hypnotics, or antipsychotic medications such as such as benzodiazepines, zolpidem, or alpram) that could influence the results of this study, and (e) participation in regular exercise programs within the last six months. Table 1 shows study participant characteristics.

2.2. Materials

2.2.1. GAITRite® System

Gait parameters were measured using the GAITRite® System (CIR Systems; Easton, PA, USA), a valid and reliable tool for measuring temporal and spatial gait parameters [20]. The active area of the system was 61 cm wide $\times$ 366 cm long with a total of 13,824 sensors placed 1.27-cm apart and covered with a roll-up carpet. The active measurement area was activated by mechanical foot pressure on the mat. Data from the activated sensors were collected by a computer at a sampling rate of 80 Hz, and the gait parameters were automatically identified and calculated.

2.2.2. Fitmeter accelerometer

A tri-axial accelerometer (Fit Dot Life; KOR) that was $35 \times 35 \times 13$ mm and weighed 13.7 g was used to measure COM acceleration during walking under three different conditions. The range of the sensors was ± 2 to ± 8 g, and the precise value was selected with the acquisition software (Fitmeter Manager 2, ver. 1.2.0.14; KOR); a range of ± 2 g was selected for the present study. The accelerometer was fixed with double-sided adhesive tape over the L3 spinous process [21]. Another accelerometer with a hand switch was used by the investigator for measuring into start and finish time point of task. The raw data were measured using x, y, and z acceleration variables. The data were automatically transferred to a computer using a USB cable connection. The COM acceleration trajectory was calculated using a two-point finite difference method [22]:

$$\text{COM acceleration} = \sum_{i=1}^{n-1} \sqrt{(x_{i+1}[\![- x_i)]\!]^2 + (y_{i+1}[\![- y_i)]\!]^2 + (z_{i+1}[\![- z_i)]\!]^2}$$

Data were collected with a sampling rate of 32 Hz.

Table 1
Characteristics of the participants.

Variables	PBVA group ($n = 9$)	GBVA group ($n = 10$)	P
Age (years)	75.67 ± 6.33	76.40 ± 7.04	0.815
Height (cm)	146.69 ± 2.72	149.19 ± 3.53	0.104
Body weight (kg)	49.18 ± 5.24	50.69 ± 4.34	0.501
VA (left side)	0.56 ± 0.11	0.23 ± 0.07	0.000*
VA (right side)	0.56 ± 0.13	0.21 ± 0.09	0.000*

All values are mean ±standard deviation.
Abbreviation: PBVA: poor binocular visual acuity; GBVA: good binocular visual acuity; BVA: binocular visual acuity (logMAR).
* $P < 0.05$.