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Does arm swing emphasized deliberately increase the trunk stability during walking in the elderly adults?

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ABSTRACT

The purpose of this study was to determine whether trunk stability while walking changes when arm swing is deliberately altered in elderly individuals. Participants included 21 community-dwelling elderly individuals (7 men and 14 women; age, 81.8 ± 5.0 years). We measured trunk acceleration by using a wireless miniature sensor unit containing a tri-axial linear accelerometer under 3 walking conditions: normal walking (normal condition), deliberately walking without any arm swing (no swing condition), and walking with a deliberately emphasized arm swing (over swing condition). To evaluate trunk stability during walking, we calculated harmonic ratios (HRs) based on trunk tri-axial acceleration signals (anteroposterior: AP, vertical: VT, and mediolateral: ML). HR–AP and HR–VT were not significantly different across the 3 conditions, but HR–ML in the over swing condition was significantly higher than that in the other 2 conditions by generalized estimating equations (GEE) adjusted for walking speed ($p < 0.05$). These findings indicate that trunk stability in the ML direction increased when the elderly individuals walked with a deliberately emphasized arm swing.

1. Introduction

It is generally believed that maintenance of walking stability is required for successful locomotion. However, age-related decline in neuromuscular function make this difficult. Accordingly, elderly individuals compensate for their reduced physical capabilities by being more cautious and by changing their gait pattern [1]. For example, compared to young people, elderly individuals have shortened step lengths, slower walking speeds [2], increased variability in stride time [3], and decreased arm swing [4]. Despite these compensatory strategies, the elderly still exhibit an unstable walking pattern [5].

Control of trunk movement is important for the maintenance of walking stability in the elderly. Studies of upper body motion during walking have revealed that the trunk plays an important role in minimizing the motion of the head by attenuating gait-related oscillations, presumably to regulate input to the visual and vestibular apparatus. Age-related gait changes induce trunk instability, and elderly individuals who find it more difficult to control trunk stability during walking are at greater risk of falls [6]. In addition, the trunk is connected to the arms through the shoulder joints and is a key segment that drives arm movement while walking. Similarly, arm movement affects trunk movement while walking [7]. This means that trunk and arm movements interact with each other.

Several studies have shown that adequate arm swing contributes to successful postural control of the body and increased efficiency of walking. Among younger individuals, compared to normal walking, walking with restriction of arm swing induces lower transverse thoracic rotation [8], indicating that there is a strong relationship between movements of the trunk and arm while walking. Moreover, the restriction of arm swing also affects

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other gait-related characteristics; it decreases walking speed and stride length [9]. In contrast, deliberately emphasizing arm swing during walking would help improve stability during walking in young and middle-aged adults [10,11]. However, some have argued that arm swing has negative effect or no effect on stability [12,13], which indicates lack of consensus. Behrman et al. showed that verbal instructions to deliberately emphasize arm swing increased walking speed and stride length, even in healthy elderly individuals [14]. This study suggested that deliberately changing the arm swing altered the walking pattern in these individuals. However, no study to date has discussed the effects of arm swing on trunk stability while walking in elderly individuals.

Therefore, this study aimed to examine changes in walking pattern when arm swing was deliberately altered in elderly individuals. We hypothesized that trunk stability would be reduced when arm swing was deliberately restricted, and would be increased when arm swing was deliberately emphasized while walking.

2. Methods

2.1. Participants

Twenty-one community-dwelling elderly individuals (7 men and 14 women; age, 81.8 ± 5.0 years) participated in this study. Inclusion criteria were the ability to walk without assistance (use of an assisting device or support by another person), ability to follow the instructions of the investigators, and lack of any neurological disease that could influence walking, any musculoskeletal disease in the upper extremities that could affect arm swing during walking, as well as any musculoskeletal disease in the lower extremities that could affect walking. Participant characteristics were assessed using the 15-item Geriatric Depression Scale (GDS) for evaluating depressive symptoms [15], the short physical performance battery (SPPB) for evaluating lower extremity function [16], and the Tokyo Metropolitan Institute of Gerontology Index of Competence (TMIG) for evaluating functional capacities [17]. The GDS includes 15 items, and participants whose total score was $\geq$10 were defined as having depressive symptoms. The SPPB total score is 0–12, and participants whose total score was $\leq$9 were defined as having poor lower extremity function [16]. The TMIG includes 13 items and 3 subscales (instrumental activities of daily living (IADL), intellectual activity, and social role), and a lower score for this index is defined as low physical functioning [18]. A previous history of falls was assessed by using the question “Have you fallen in the last year?” with 2 response categories (yes/no). The Research Ethics Committee of the Kobe University Graduate School of Health Sciences approved this study, and written informed consent was obtained from all subjects in accordance with the Declaration of Helsinki.

2.2. Walking procedure and apparatus

The participants walked on a 20 m long smooth, horizontal walkway. The walkway was divided into 15 m by 2 lines positioned 2.5 m from each end to allow for acceleration and deceleration. Participants were verbally instructed to walk under 3 conditions: (1) normal walking at preferred, usual speed (normal; *N*), (2) deliberately walking without any arm swing (no swing; NS), and (3) walking with a deliberately emphasized arm swing (over swing; OS). The participants were also verbally instructed not to change their walking speed across the conditions in order to minimize the influence of speed changes. The acceleration and angular velocity of the trunk, heel, and arm were measured while walking by using 3 wireless miniature sensor units (MVP-RF-8, MicroStone Co., Nagano, Japan) under every condition. Each sensor unit contained a tri-axial angular rate gyroscope and a linear accelerometer. The sensor units were attached to a fixed belt at the level of the L3 spinous process, the posterior surface of the right heel, and the dorsal aspect of the right wrist by using surgical tape. All signals were sampled at 200 sample/s and were wirelessly and synchronously transferred to a PC via a bluetooth personal area network. During each trial, the participants were timed as they walked over a 10 m distance marked on the walkway, and their walking speed was expressed in meters per second. In addition, stride length was calculated by using the formula, average walking speed multiplied by the average stride duration calculated over 10 strides, according to the method reported previously [19].

2.3. Data analysis

Signal processing was performed by using MATLAB software (Release 2008b, The MathWorks, Nattick, MA). Before analysis, all acceleration and angular velocity data were low-pass filtered with a cutoff frequency of 20 Hz. Based on pilot testing to determine temporal parameters using the heel acceleration and heel angular velocity data, the heel contact event was identified by a vertical acceleration peak. Harmonic ratios (HRs), the primary dependent variables, were calculated using trunk acceleration data over 10 strides while walking in a steady state, and were expressed as an average value per stride. HRs based on trunk tri-axial acceleration signals measured trunk smoothness and rhythmicity while walking. Emerging evidence showed the usefulness of applying the HR technique by using trunk acceleration to differentiate characteristics among individuals [6,20,21]. Higher HRs correspond to greater smoothness and are indicative of trunk stability while walking, [5,22,23]. HRs were computed by using separate digital Fourier transformations for each direction (anteroposterior [HR–AP], vertical [HR–VT], and mediolateral [HR–ML]). Typical AP and VT acceleration patterns of the trunk during walking exhibit 2 major acceleration peaks per stride, one for each step. The even harmonics of the AP and VT planes indicate the in-phase components of the signal, whereas the odd harmonics comprise the out-of-phase components (minimized in healthy gait). As a result, HR–AP and HR–VT were calculated by dividing the even harmonics (summed amplitudes of the first 10 even harmonics) by the odd harmonics (summed amplitudes of the first 10 odd harmonics). Conversely, the basic ML pattern is limb dependent and only repeated once for any given stride. Consequently, ML accelerations are dominated by odd rather than even harmonics. Therefore, HR–ML was calculated from a ratio of the odd harmonics divided by the even harmonics [5,23]. Further, to quantify the arm swing during each walking condition, the angular velocity data for the sagittal plane from the right wrist during steady state walking were integrated for each stride, and the average value of the angular variation for each step was computed.

2.4. Statistical analysis

All analyses were performed using SPSS 21.0 J for Windows (SPSS Japan Inc., Tokyo, Japan). Mean arm swing, stride length, and walking speed were compared across walking conditions using a univariate generalized estimating equation (GEE). Since repeated measurements of a subject are dependent on each other, correction is needed for these within-subject correlations. Using the GEE, this correction was conducted by the addition of a correlation structure as a covariate to the analysis. In these analyses, an exchangeable working correlation structure and robust estimation of the covariance matrix were used. Additionally, since changes in the walking speed of the subjects would influence the HRs, we considered it a confounding factor and used the GEE analyses adjusted for walking speed to compare HRs across walking