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ANKLE DORSIFLEXOR STRENGTH RELATES TO THE ABILITY TO RESTORE BALANCE DURING A BACKWARD SUPPORT SURFACE TRANSLATION

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Abstract

Functional base of support (FBOS), the effective area for center of pressure (COP) movement, decreases with aging, which would reduce one's ability to restore balance during perturbed stance. We investigated the relationship between ankle muscle strength and FBOS as well as the threshold perturbation acceleration that required a heel-rise (HR) or step (STEP) to maintain balance. Standing posture of 16 young and 16 elderly adults was perturbed with a backward support surface translation with the speed ranging from 15 to 70cm/s. Dorsiflexor (DF) strength was found to significantly correlate with FBOS measures and threshold acceleration for HR. Significant correlations were also found between FBOS measures and threshold accelerations for HR and STEP, except for the backward FBOS and threshold acceleration for STEP. Elderly subjects demonstrated significantly smaller DF strength and FBOS measures than young subjects, but no significant group difference was detected in plantarflexor (PF) strength. Most elderly subjects took a step once they raised their heels, while most young subjects were able to restore balance after heel-rise. These findings, taken together, imply that weakness in ankle dorsiflexors could limit the ability of elderly adults to restore balance while standing on their toes. FBOS measures and ankle dorsiflexor strength could be sensitive measures to detect individuals with declined balance control.

Keywords

Balance; Perturbation; Functional base of support; Ankle muscle strength

1. Introduction

Falls are one of the most serious problems among the elderly, resulting in fatal physical injuries [1–3]. Poor postural control is one of the contributing factors to increased incidence of falls in the elderly [4]. Responses to unexpected perturbations during standing, such as support surface translations, have been commonly used to examine dynamic postural control

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Conflict of interest

The authors have no conflicts of interest in relation to the work reported here.

[5–9]. Elderly adults are reported to take compensatory steps following a smaller balance perturbation than that which requires young adults to take steps [5, 8]. Occurrence of a stepping response has been traditionally regarded as a consequence of the center of mass (COM) moving beyond limits of the base of support (BOS) [10]. However, even with a location inside the BOS, a standing posture might not be maintained if the COM traverses with a sufficiently large horizontal velocity [5, 7, 8, 11]. Therefore, step initiation could depend on the COM position in relation to the BOS and its instantaneous velocity, i.e., momentum.

One possible explanation for increased incidence of stepping in elderly adults would be an inability to generate the necessary muscle torques to control the horizontal COM momentum [10]. However, no significant differences between young and healthy elderly adults were found for the maximum ankle muscle torque achieved prior to step initiation or rate of torque development following the onset of perturbation [10, 12], suggesting that neither the magnitude nor rate of ankle muscle torque production are limiting factors in the increased incidence of stepping in the elderly.

Although muscle torque responses for momentum control may not be different between young and elderly adults, differences could be found in the effective limits of the BOS, which provides stability margins for the COM movement. To maintain balance during support surface translations, actively controlling the center of pressure (COP) is required to keep the COM within a desired region, as the distance between the COP and COM correlates with the horizontal COM acceleration [13], which regulates momentum. The BOS could serve as the limits for COM control since it provides a possible range for COP movement. It has been shown that the functional base of support (FBOS), defined as the effective limits of COP movement, decreases with aging [14]. A decrease in FBOS indicates a constriction of the limits of stability for COM control, which would reduce an individual's ability to restore balance.

Studies of quiet or perturbed standing have reported the dominance of ankle muscles in balance maintenance in the antero-posterior direction [15]. It has been suggested that reduced ankle muscle strength is a contributing factor to the loss of balance in elderly adults [12, 16], and enhancement of ankle muscle strength could lead to improvements in balance recovery during standing perturbations in the elderly [17]. Since FBOS is determined using COP excursions during sustained maximal leaning about the ankle joints, weakness in ankle muscles would contribute to a decreased FBOS and explain the increased incidence of stepping in the elderly when stance is perturbed.

The objective of the study was, therefore, to examine the relationship between ankle muscle strength, FBOS, and the ability to maintain balance, which was assessed as the threshold perturbation acceleration that required a heel-rise or step response. We hypothesized that FBOS would be a sensitive measure to predict threshold perturbation acceleration for taking a step or heel-rise, and ankle plantarflexor strength would be a significant predictor for FBOS as well as threshold perturbation acceleration during a backward support surface translation. It was also hypothesized that significant reductions in ankle muscle strength and FBOS would be observed in the elderly who could sustain smaller threshold perturbation accelerations, while no differences would be observed in ankle torque generation rate between young and elderly subjects.

2. Methods

Standing posture of 16 healthy young adults [8 men/8 women; mean age 25.4 ± 4.3 years, mean height 171.3 ± 8.7 cm, mean mass 67.7 ± 12.5 kg] and 16 healthy elderly adults [7 men/9

women; mean age 74.9 ± 6.2 years, mean height 163.1 ± 9.7 cm, mean mass 67.5 ± 10.4 kg] was perturbed with a backward support surface translation. Subjects stood with one foot on each of two electronically synchronized force platforms (Institute of Neuroscience, University of Oregon) with their arms folded on their chest [18]. They wore a lightweight harness attached to an overhead trolley to ensure safety and an assistant remained by the side of the subject to prevent a fall. Subjects were instructed to maintain their balance without taking a step and not to bend their trunk with hip flexion. Six perturbations included one for each of the velocities 15, 30, 40, 50, 60, and 70 cm/s and the platform moved backward for 15 cm at all velocities. Platform displacement was achieved with nonlinear ramp-to-parabola acceleration waveforms, which included a ramp onset for acceleration followed by a parabolic offset for deceleration [18]. Perturbations were presented in a pseudo-randomized manner whereby less severe disturbances occurred within earlier trials [10]. This testing order served as a precautionary measure to ensure that subjects would not be exposed to the more severe disturbances early in the test session. The experimental protocol was approved by the Institutional Review Board. No participants had a history or clinical evidence of neurological, musculoskeletal or other medical conditions.

Motion data were captured with an eight-camera motion analysis system (Motion Analysis Corp., Santa Rosa, CA) at 120Hz. A total of 29 reflective markers were placed on each subject's bony landmarks [19]. Additionally, 4 markers were placed on the corners of the moving platform to obtain platform movement. Marker trajectory data were low-pass filtered using a fourth-order Butterworth filter with a cut-off frequency of 8Hz. Perturbation acceleration was used to describe the magnitude of the balance disturbance since previous studies have shown that platform acceleration provides an initial destabilizing input in the postural response [18, 20–24]. Perturbation acceleration for each speed condition was calculated from the 2nd derivative of plate marker displacement. The onset of platform motion was defined as the first zero crossing in a backward acceleration phase of the platform acceleration (Fig.1) [18]. Ground reaction forces were collected at 960 Hz. Platform data from unloaded trials were collected at each testing session and were subtracted from loaded trials to eliminate movement artifacts in force measurements [25]. Responses to the platform perturbation were classified as in-place, step (STEP), and/or heel-rise (HR). A STEP response occurred if a step was observed. A HR response occurred if heels were raised more than 1% of body height. An in-place response occurred when the feet remained in contact with the supporting surface throughout the trial with heels raised less than 1% of body height. HR responses included STEP responses when an initial HR response was followed by a STEP response.

Electromyographic (EMG) data were collected to identify onset of muscle activities, which was later used to determine the active phase of ankle torque development. Bipolar surface electrodes (Motion Lab Systems Inc., Baton Rouge, LA) were placed on the skin over the right lateral gastrocnemius (GA) and sampled at 960 Hz. EMG data were rectified, but not filtered for analysis to determine muscle onset latencies. Criteria for determining muscle onset latencies were that EMG activity was greater than the baseline mean plus three standard deviations, and that half of the EMG sample points in the burst remained above this level for at least 40 ms [12]. Baseline data for each trial were collected 2 to 3 seconds prior to onset of plate movement.

COP positions in the antero-posterior (AP) direction during sustained maximal forward and backward leaning were calculated to determine forward FBOS (FFBOS), backward FBOS (BFBOS), and total FBOS (TFBOS) [14]. FFBOS and BFBOS were calculated from the ankle joint with the forward direction as positive, and TFBOS was the total length between forward (FFBOS) and backward (BFBOS) limits. Ankle plantar and dorsi-flexor strengths (PF and DF, respectively) of the dominant leg were measured during isometric maximum

voluntary contraction in a seated position at a neutral ankle position with a dynamometer (Biodex Medical Systems, NY) and were normalized to body mass.

The combined sagittal plane joint moment from both ankles during each perturbation trial was calculated based on a foot segment model using collected ground reaction forces and kinematic data [4]. Anthropometric data were estimated using the initial work of Dempster [26]. Peak ankle joint moment, normalized to body mass, was defined as the maximum ankle torque following the onset of plate movement. If a step occurred during the trial, the maximum torque achieved prior to step onset was chosen. Step onset was defined as the instant when the foot was lifted from the floor and was detected using the ground reaction force. The active phase of ankle torque development (torque generation rate) was determined as the rate of ankle joint moment generation over a period of 60 ms following the onset of GA muscle activity [12]. If a stepping event took place during this time period, this rate of ankle joint moment generation was not calculated [12]. The peak COP displacement was calculated as the largest distance between the COP and ankle during each trial (or until step onset).

An independent *t*-test was performed to examine group differences in ankle muscle strength, FBOS measures, muscle onset latency, ankle torque generation rate, peak ankle joint moment, and peak COP displacement. Significance level was set at $\alpha = .05$. Pairwise comparisons were analyzed with adjustments for multiple comparisons using Bonferroni procedure ($p_{\text{PC}} = .05/6 = .0083$). Linear regression analyses were performed to examine the relationship between ankle muscle strength, FBOS measures, and threshold perturbation accelerations for STEP and HR responses (TAccSTEP and TAccHR, respectively). Threshold perturbation acceleration was the lowest platform acceleration that a STEP or HR response was observed for each subject.

3. Results

Overall, 19.7% and 65.5% of the overall number of trials were STEP responses, and 75.2% and 88.5% were HR responses for young and elderly subjects, respectively. In addition, 26.2% and 74.0% of HR responses were followed by STEP responses for young and elderly subjects, respectively. As the perturbation speed increased, the number of STEP and HR responses increased, except for the fastest perturbation condition for STEP response (70cm/s) (Fig.2).

For all subjects, DF strength was found to be significantly correlated with FFBOS, BFBOS, TFBOS, and TAccHR, accounting for 24~41% of the variance (Table 1). Significant correlations were also found between all the FBOS measures and threshold accelerations for HR and STEP, except for BFBOS and threshold acceleration for STEP response. Within each subject group, significant correlation was detected only between DF strength and TAccHR for the young group.

Elderly subjects showed a significantly smaller DF strength than young subjects ($p < .001$), but no significant group difference was found in the PF strength. Elderly subjects also showed significantly smaller FFBOS, BFBOS, and TFBOS ($p > .006$, Table 2). No significant differences were detected in either muscle onset latency or ankle torque generation rate between young and elderly subjects for all perturbation conditions (Fig.3a and 3b). However, elderly subjects demonstrated significantly smaller peak ankle joint moments and COP displacements ($p = .008$ and $p = .004$, respectively) than young subjects in the 60 cm/s perturbation speed condition (Fig.3c and 3d).

4. Discussion

The objective of this study was to examine the relationship between ankle muscle strength, FBOS, and the threshold perturbation acceleration that required a heel-rise or step to recover balance from a backward support surface translation. Contrary to our hypothesis that the plantarflexor strength would be a significant predictor of FBOS and threshold perturbation acceleration, it was the dorsiflexor strength that significantly correlated with FBOS measures and threshold perturbation accelerations that required heel-rise responses. Individuals with weaker dorsiflexor strength showed smaller FBOS measures and raised their heels during backward perturbations with smaller accelerations.

Higher rates of STEP and HR responses in elderly compared to young subjects were observed for each of the perturbation speeds. As the perturbation speed increased, the number of STEP and HR responses increased for both young and elderly subjects, except for the fastest perturbation condition for STEP responses. This is because platform perturbations were presented in a pseudo-randomized manner whereby slower translations occurred first. Some subjects who took a step during these earlier trials were not tested for the fastest perturbation condition for safety purposes, which resulted in a decrease in percent STEP responses in the fastest perturbation condition. Although it has been reported that elderly adults take a compensatory step with smaller perturbation magnitudes than young adults [5–9], our findings further revealed that most elderly subjects took a step once they raised heels (74.0% of HR responses), while most young subjects were able to restore their balance after heel-rise without stepping (73.8% of HR responses). Backward translations were reported to induce muscle activities in the tibialis anterior as individuals are standing on their toes, suggesting that balance recovery after heel-rise could require dorsiflexor activation [27]. A significant reduction in dorsiflexor strength, as observed in elderly subjects, could contribute to the reduced ability to recover balance from backward perturbations.

No significant differences between the young and elderly were detected in either muscle onset latency or ankle torque generation rate in response to perturbations, which is in agreement with previous findings [12]. Peak ankle joint moments of elderly subjects were found to be smaller for the faster perturbation speeds when compared to young subjects. Although a significant group difference was detected only at the 60cm/s condition, the same trend was observed in other perturbation speeds (30cm/s: $p=.028$; 40cm/s: $p=.018$; 50cm/s: $p=.014$; 70cm/s: $p=.014$). These results supported that ankle torque generation rate following a backward perturbation may not be a contributing factor to increased incidence of stepping in the elderly, but suggested that peak ankle joint moment could still be a limiting factor.

Elderly subjects showed a significantly smaller FBOS, which indicates a reduction in the effective limits of COP movement. Smaller peak COP displacements observed in elderly subjects were found to be consistent with this reduction as FBOS confines the functional boundary in which the COP could be displaced. The smaller peak ankle joint moments obtained in elderly subjects could explain this result, since the ankle joint moment is proportional to the distance between the COP and the ankle joint. Moreover, significant correlations were detected between FBOS measures and threshold perturbation accelerations. FBOS measures seem to be a sensitive measure to predict threshold perturbation accelerations for taking a step or heel-rise. The increased rate of stepping in the elderly during backward perturbations could be due to a reduced FBOS, which limits the range of COP movement to control the COM to achieve upright standing.

A possible reason for a reduced FBOS in elderly adults could be their significantly weaker dorsiflexor strength as compared to young adults. Individuals with weaker dorsiflexor strength showed smaller FBOS measures and raised their heels with smaller perturbation

speeds, while plantarflexor strength did not correlate with any of these measures. Age-related declines in the ankle dorsiflexor strength have been demonstrated previously, where significant differences were detected between healthy older adults and fallers [16]. Furthermore, a successful balance recovery following a heel-rise would require dorsiflexor activation [27]. This could explain that most elderly subjects in the present study took a step once they raised their heels, while most young subjects did not. It has also been shown that increased muscle activation on the anterior aspect of the body was observed as the velocity of backward translations increased [28]. Dorsiflexor strength was reported to be significantly associated with the maximum recoverable forward lean angle of older adults [29], in support of our findings that the forward FBOS was significantly associated with ankle dorsiflexor strength. These findings, taken together, suggest that responses from ankle dorsiflexors could also be important for balance maintenance during backward perturbations that induce a forward body sway. Elderly adults may not be able to control the COP and maintain balance while standing on their toes as effectively as young adults due to weakness in ankle dorsiflexors, which would limit their ability to restore balance from a backward support surface translation.

There are several limitations in this study. Although no significant differences were detected in plantarflexor strength between young and elderly subjects, elderly subjects showed smaller peak ankle plantarflexor moments during the faster perturbations. Plantarflexor muscles are activated eccentrically to restore balance during a forward body sway induced by a backward perturbation. As the magnitude of muscle force during an eccentric activity is known to be stronger than that of an isometric activity [30], the plantarflexor strength measured in an isometric and non-weight bearing condition might not reflect the maximum torque production during dynamic tasks. It should also be noted that the group difference in the peak ankle plantarflexor moment was detected with the data combined from all perturbation responses (in-place, HR, and STEP). Different rates observed between young and elderly subjects in each perturbation response might confound this finding. However, we were not able to systematically examine the group differences in the peak ankle plantarflexor moment for each of the three responses across all perturbation speeds due to insufficient or no data available for some responses and/or speed conditions. Nevertheless, with the data available, no significant effects of response types were detected on the peak ankle moments ($p = .37$) for each subject group. In addition, despite our instruction asking subjects not to bend their trunk with hip flexion, hip flexion angles were still observed. However, no significant differences in hip flexion angle were detected between young and elderly subjects for all perturbation conditions (Young: $21 \pm 9^\circ$; Elderly: $18 \pm 8^\circ$).

In conclusion, ankle dorsiflexor strength was found to be significantly associated with FBOS measures as well as threshold perturbation speed for heel-rise during backward platform translations. Elderly subjects demonstrated a decreased FBOS, which would reduce the ability to recover balance from backward support surface translations. FBOS measures and ankle dorsiflexor strength could be sensitive measures to detect elderly individuals with a decline in balance control.

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Highlights

We examined the threshold perturbation acceleration to maintain balance.

Most elderly subjects took a step once they raised their heels.

Weakness in ankle dorsiflexors limits ability of elderly adults to restore balance.

Fig.1. Representative plate displacement and acceleration profiles for the 40cm/s condition. The onset of platform motion was defined as the first zero crossing in the backward acceleration phase of the platform acceleration.

(a) Young

(b) Elderly

Fig.2.
Number of STEP, HR, and In-place responses in percentage by perturbation speed for (a) Young and (b) Elderly.

Fig.3.

Group comparisons for (a) muscle onset latency, (b) ankle torque generation rate, (c) peak ankle joint moment, and (d) peak COP displacement normalized by foot length (†, * $p < .0083$)

Table 1

Results of linear regression analyses (PF and DF: Plantarflexor and Dorsiflexor strengths, respectively, FFBOS: Forward FBOS, BFBOS: Backward FBOS, TFBOS: Total FBOS, TAccSTEP and TAccHR: Threshold perturbation accelerations for STEP and HR responses, respectively).

Predicted variables	Young		Elderly		Overall	
	Regression equation	R ²	Regression equation	R ²	Regression equation	R ²
FFBOS[m]	-0.001PF+0.164 0.041DF+0.144	0.00 0.10	0.011PF+0.131 0.067DF+0.122	0.04 0.07	0.009PF+0.144 0.083DF*+0.120	0.02 0.24 *
BFBOS[m]	-0.014PF+0.01 -0.032DF+0.005	0.10 0.07	0.007PF -0.033DF+0.017	0.08 0.10	-0.002PF -0.060DF*+0.022	0.00 0.29 *
TFBOS[m]	0.013PF+0.159 0.074DF+0.138	0.05 0.19	0.004PF+0.132 0.100DF+0.105	0.00 0.14	0.010PF+0.144 0.143DF*+0.099	0.02 0.41 *
TAccSTEP [m/s ²]	-1.35PF+7.97 -11.3DF+10.8 33.0FFBOS+1.3 35.2BFBOS+6.7 18.3TFBOS+3.5	0.13 0.24 0.20 0.14 0.04	0.55PF+4.30 1.34DF+4.45 25.6FFBOS+1.2 -35.9BFBOS+5.1 27.6TFBOS+1.1	0.02 0.01 0.17 0.06 0.22	0.42PF+4.89 3.48DF+4.17 34.0FFBOS*+0.4 -39.2BFBOS+5.4 35.0TFBOS*+0.3	0.01 0.04 0.23 * 0.09 0.32 *
TAccHR [m/s ²]	-1.22PF+6.00 6.09DF*+1.78 23.3FFBOS+0.8 -8.86BFBOS+4.55 18.6TFBOS+1.4	0.01 0.27 * 0.07 0.01 0.07	-0.064PF+3.885 2.78DF+2.94 9.11FFBOS+2.52 -34.6BFBOS+4.0 13.3TFBOS+2.0	0.00 0.09 0.06 0.16 0.15	-0.31PF+4.55 4.90DF*+2.31 19.1FFBOS*+1.3 -29.3BFBOS*+4.2 18.2TFBOS*+1.4	0.01 0.29 * 0.13 * 0.13 * 0.20 *

* $p < .05$.

Table 2

Ankle muscle strength (PF: Plantarflexor, DF: Dorsiflexor) and FBOS measures normalized by foot length (FFBOS: Forward FBOS, BFBOS: Backward FBOS, TFBOS: Total FBOS). A negative value for BFBOS for Young subjects indicates a larger FBOS in the backward direction since it was calculated from the ankle joint with the forward direction as positive.

Variables	Group		<i>p</i>
	Young	Elderly	
PF[Nm/kg]	1.10±0.30	1.06±0.40	.761
DF[Nm/kg]	0.47±0.11	0.31±0.10	<.001 [*]
FFBOS[%FL]	65.3±5.1	58.6±9.0	.006 [*]
BFBOS[%FL]	-3.7±5.0	2.5±4.0	.001 [*]
TFBOS[%FL]	69.0±4.0	56.0±9.7	<.001 [*]

^{*}
p < .05.