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Increased muscular challenge in older adults during obstructed gait

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Abstract

Skeletal muscle strength is known to decline with age. Although lower extremity (LE) muscle strength is critical to maintaining dynamic stability, few studies have investigated lower extremity muscle challenge during activities of daily living. The purpose of this study was to investigate the effects of age and obstructed gait on relative lower extremity muscular challenge, with respect to available joint strength. Fifteen healthy young and fifteen healthy older adults were asked to walk over level ground and step over obstacles. Pre-amplified surface electrodes were used to measure bilateral muscular activation of the gluteus medius (GM), vastus lateralis (VL), and gastrocnemius (GA). Muscle activation signals were normalized to peak magnitudes collected during maximal manual muscle testing (MMT). Normalized magnitudes were analyzed during the double-support phase for gluteus medius and vastus lateralis and during the single-support phase for gastrocnemius. A two-factor ANOVA was used to test for age group effect, with repeated measure of obstacle height. In general, older adults demonstrated greater relative activation levels compared to young adults. Gluteus medius activity was significantly greater in the elderly as compared to young during periods of double-support (weight transfer). Increased obstacle height resulted in greater relative activation in all muscles, confirming the increased challenge to the musculo-skeletal system. While healthy elderly adults were able to successfully negotiate obstacles of different heights during walking, their muscular strength capacity was significantly lower than young adults, resulting in relatively higher muscular demands. The resulting potential for muscular fatigue during locomotion may place individuals at higher risk for trips and/or falls.

Keywords: Elderly; EMG; Joint strength; Obstacle crossing; Balance control

1. Introduction

Skeletal muscle strength is known to decline with age [1,2]. As an intrinsic risk factor contributing to falls, lower extremity (LE) muscle strength is a critical component in limiting an elderly individual's dynamic stability [3–5]. It has been reported that fallers demonstrated only 37% of the knee extensor strength, and 10% of the ankle plantar flexor strength exhibited by their non-falling peers [6]. Stepping over obstacles has been shown to require greater muscle force than level walking [7], and recent results have shown significant associations between isometric strength and the ability of elderly individuals to cross obstacles [8]. Involvement in a long-term lower extremity resistance training regimen resulted in substantial strength gains among older adults (197–285% increase), along with significant improvement in obstructed gait function (speed of crossing stride, increased obstacle clearance, etc.) [9]. Additionally, it has been demonstrated that lower extremity joint strength affects stepping speed and toe trajectory during early swing [10].

Quantification of the neuromuscular challenge encountered by lower extremity muscles of healthy elderly individuals can provide baseline information against which strength declines or strength training interventions may be compared. It is interesting to note a lack of studies addressing the validity of using normalized electromyography (N-EMG) to identify age-related differences in relative levels of muscular challenge encountered during activities of daily living. This may be due in part to the non-linear EMG/force relationships reported by various groups

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[11–13]. These studies reported non-linear patterns in the EMG/force ratio for isometric contractions, but did not test EMG/force relationships during anisometric contractions. One recent study demonstrated non-linear relationships between force and EMG during anisometric contractions for the first dorsal inter-osseus muscle [14]. Their findings indicate that less voluntary muscular activation is needed during eccentric contractions as compared to concentric contractions to produce the same force.

Previous studies have reported that older adults adopt a conservative strategy when crossing obstacles [15,16], as indicated by slower crossing speed, shorter step length, shorter obstacle–heel-strike distance [15], and reduced anterior/posterior separation between whole body center of mass (COM) and the center of pressure (COP) [16]. The selection of a conservative obstacle-crossing strategy may be related to natural age-related strength loss. If so, then it follows that the relative challenges of obstacle crossing (and other functional tasks) may tax the available joint strength to a point where a person's ability to control balance dynamically could be seriously compromised. Quantification of the relative challenge imposed on lower extremity muscles would be beneficial to understanding the thresholds of joint strength that are needed to allow adequate negotiation of obstacles in daily life.

The purpose of this study was, therefore, to investigate the effects of age and obstacle height on relative lower extremity muscular challenge, with respect to available joint strength. It was hypothesized that healthy elderly adults would require a greater percentage of their neuromuscular capacity (as measured by increased N-EMG values) during level walking and while crossing an obstacle. We further hypothesized that N-EMG levels increase as obstacle height increases; indicating the task-specific challenge imposed on lower extremity muscles during this functional activity.

2. Methods

Fifteen young adults (8 males/7 females; 24.5 ± 3.6 years, 172.3 ± 6.8 cm, 72.5 ± 10.1 kg) and 15 elderly adults (9 males/6 females; 72.6 ± 5.5 years, 168.3 ± 9.5 cm, 75.8 ± 12.2 kg) were recruited for this study from the University of Oregon campus and the surrounding community, within the guidelines of the Institutional Review Board. Inclusion criteria required that subjects had no histories of significant head trauma, neurological disease (e.g. Parkinson's, post-polio syndrome, diabetic neuropathy), visual impairment not correctable with lenses, musculo-skeletal impairments (e.g. amputation, joint replacement, joint fusions, joint deformity due to rheumatoid arthritis), or persistent symptoms of vertigo, light-headedness, unsteadiness. All of the subjects were community-dwelling individuals. Elderly subjects were noted to be active community members, with many of them currently involved in recreational sporting activities.

Pre-amplified surface electrodes were attached bilaterally over the bellies of the gluteus medius (GM), vastus lateralis (VL), and medial head of the gastrocnemius (GA). These muscle groups were previously shown to be substantially challenged when stepping over obstacles [17,18]. Activation magnitude of each muscle during gait was normalized to values taken during maximal effort manual muscle testing (MMT). Maximal GM activation was tested in 30° of hip abduction, while side lying. For VL maximum, subjects were seated with the knee in 45° of flexion. Maximal GA activation was tested in neutral ankle position, with the subject fixed to a table in prone position. MMT procedures were performed by one examiner for each muscle group, bilaterally. Subjects were verbally encouraged to ensure maximal recruitment.

Subjects were then asked to walk at a self-selected pace during level and obstructed gait trials. Level walking trials were performed first, followed by obstacle crossing trials. A single obstacle consisting of two upright standards and a light-weight crossbar was set to four height conditions; 2.5%, 5%, 10%, and 15% body height (BH). The relative difficulty of stepping over obstacles was thus normalized to individual body height, accounting for variation within the sample. This obstacle crossing protocol has been accepted previously [19–21]. Obstacle heights were randomized with three trials collected for each condition. The leading limb was defined as the first limb to cross the obstacle. Crossing stride was defined as the trailing limb heel-strike before the obstacle to heel-strike of the same limb after crossing the obstacle.

For all MMT and gait trials, raw EMG signals were collected at 960 Hz using the MA-300™ system (Motion Lab Systems, Inc., Baton Rouge, LA), band-pass filtered (20–350 Hz), full wave rectified, and passed through a linear envelope at 10 Hz for final interpretation. Filtered signals from the gait trials were then normalized to the MMT signal maximum for each muscle to indicate relative activation levels. In this way, the relative activation values were recorded as a percentage of the maximum activation available to each individual muscle (N-EMG). Fig. 1 demonstrates the data-processing steps leading from full-wave rectified EMG to smoothed EMG data (time-normalized to 100% crossing stride). The mean value (within-trial) for each support phase in the gait cycle (double-support and single-support) was calculated for the leading and trailing limb.

Initial statistical assessment required screening of outliers using inter-quartile range. If a case was found to be more than three times the inter-quartile range away from the median, that case was removed from further analysis. Mean N-EMG values were analyzed for the effects of age group and obstacle height (two-factor ANOVA with repeated measures of obstacle height). Significance level was set at $\alpha = 0.05$ for all tests. Statistical analyses were conducted with SYSTAT (Version 9, SPSS Inc., Chicago, IL).

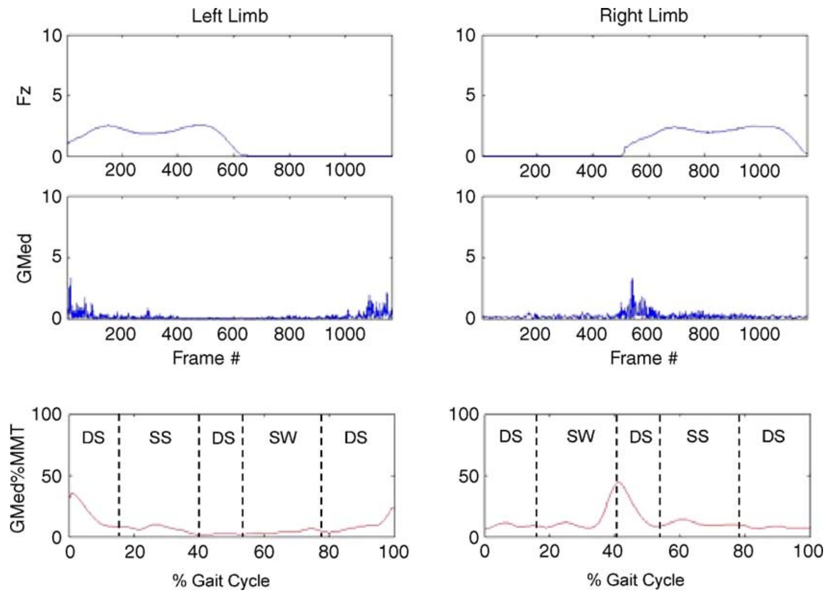

Fig. 1. Representative level walking condition showing vertical ground reaction force patterns from consecutive foot strikes, rectified EMG activation of the GM, and smoothed GM activation (after normalization to MMT). The time scale of the normalized EMG patterns is with respect to the crossing stride. Support and swing phases of gait are indicated: DS – double-support, SS – single-support, SW – swing. F_z represents the vertical ground reaction force.

3. Results

Initial screening indicated that eight data points were outside reasonable variability (±3 inter-quartile ranges). Of the 2700 total data points in the analysis (6 variables × 30 subjects × 5 conditions × 3 trials), the outlying data accounted for only 0.3% of the total data collected. Removal of these data from the analysis was, therefore, felt to be justified.

All subjects were able to complete the required measures of MMT and gait analysis trials. No incidents of tripping were observed. The testing session typically required 2 h for completion. Subjects did not indicate discomfort during any of the testing conditions, nor did they express any sense of fatigue at the end of the testing session.

No significant age group differences were found for any of the temporal-distance parameters (Table 1). As obstacle height increased so did stride time ($p < 0.001$) and stride length ($p = 0.005$). Gait velocity was found to decrease linearly with increased obstacle height ($p < 0.001$). Since the gait velocity showed no significant difference between age groups ($p = 0.182$), it was not entered as a covariate in the analysis of following N-EMG values.

Healthy elderly adults showed greater relative activation levels in the leading and trailing limbs, compared to young adults (Table 2). During double-support, weight transfer and acceptance occurs laterally as well as anteriorly. In this phase, the GM of the healthy elderly was activated up to an average of 46% of their maximum capacity, compared to 23% in the young for all testing conditions. Similarly, the VL of the healthy elderly was activated up to an average of 35% of their capacity in double-support phase, compared to 25% in the young. Maintenance of dynamic stability and forward progression is required during the single-support phase of gait. During single-support, healthy elderly GA activity reached 45% of MMT

Table 1
Temporal-distance variables compared between groups and across the obstacle heights: group mean (S.D.)

Obstacle height	None		2.5%		5%		10%		15%		p-Values[a]
	Young	Elderly	Young	Elderly	Young	Elderly	Young	Elderly	Young	Elderly	
Gait velocity (m/s)	1.363 (0.169)	1.295 (0.097)	1.331 (0.161)	1.268 (0.148)	1.317 (0.147)	1.237 (0.141)	1.253 (0.155)	1.169 (0.183)	1.195 (0.140)	1.120 (0.219)	$P_h < 0.001$, $P_g = 0.182$
Stride time (s)	1.032 (0.072)	1.008 (0.118)	1.098 (0.086)	1.071 (0.163)	1.118 (0.082)	1.097 (0.166)	1.185 (0.117)	1.140 (0.196)	1.244 (0.121)	1.200 (0.221)	$P_h < 0.001$, $P_g = 0.544$
Stride length (cm)	140.0 (13.84)	135.4 (11.69)	145.4 (14.49)	139.8 (14.62)	146.6 (13.84)	139.4 (13.32)	147.4 (14.72)	140.0 (20.68)	147.7 (13.84)	139.5 (22.82)	$P_h = 0.005$, $P_g = 0.265$

[a] P_h represents height effect. P_g represents group effect.

Table 2
Normalized EMG activation percentages during double-support phases for GM and VL, single-support phases for GA: group mean (S.D.)

Limb/muscle	Age group	Obstacle height (% BH)					
		Level	2.5	5.0	10.0	15.0	Effect
Tr. GM	Elderly	38.03 (13.65)	39.41 (11.34)	43.54 (16.21)	48.48 (19.13)	49.05 (14.26)	a,b
	Young	21.91 (12.64)	22.94 (12.76)	23.83 (12.15)	25.73 (12.90)	28.44 (14.44)	
Ld. GM	Elderly	40.24 (19.13)	45.13 (31.35)	53.26 (35.33)	50.44 (30.67)	52.76 (29.86)	a,b
	Young	19.74 (11.81)	20.67 (9.06)	21.42 (10.35)	22.45 (8.95)	25.28 (10.73)	
Tr. VL	Elderly	35.56 (18.95)	34.00 (14.94)	33.72 (15.31)	34.70 (14.73)	35.46 (15.12)	b
	Young	22.28 (12.16)	25.35 (13.16)	25.83 (13.05)	26.13 (13.08)	27.61 (11.69)	
Ld. VL	Elderly	28.89 (13.05)	35.96 (20.50)	33.90 (15.33)	37.84 (17.52)	39.00 (15.81)	a,b
	Young	20.39 (12.84)	24.05 (13.30)	24.46 (14.37)	25.27 (14.38)	25.36 (12.06)	
Tr. GA	Elderly	40.93 (18.80)	51.47 (34.32)	46.01 (25.42)	47.40 (21.54)	51.41 (30.73)	b
	Young	31.48 (13.57)	32.55 (15.14)	33.08 (14.48)	34.23 (15.78)	39.10 (14.88)	
Ld. GA	Elderly	38.40 (14.15)	45.31 (35.06)	39.11 (18.57)	40.38 (18.86)	48.42 (33.00)	b
	Young	35.50 (13.10)	35.88 (12.80)	39.57 (14.39)	40.12 (15.06)	41.18 (16.59)	

Tr.: trailing limb; Ld.: leading limb.
[a] Significant age group effect ($p < 0.05$).
[b] Significant obstacle height effect ($p < 0.05$).

maximum for all testing conditions, while the young required 36% of their capacity.

For the trailing limb, there was a significant age effect on the normalized activation levels of the GM ($p = 0.003$; Fig. 2), but not the VL ($p = 0.053$) or the GA ($p = 0.360$). In the leading limb, significant age effects were found for the GM ($p < 0.001$) and VL ($p = 0.042$), but not for the GA ($p = 0.154$). Increased obstacle height resulted in an increased relative activation of all muscles of both the leading and trailing limb ($p \leq 0.018$). Results of the ANOVA revealed no significant interactions between the factors of group and obstacle height.

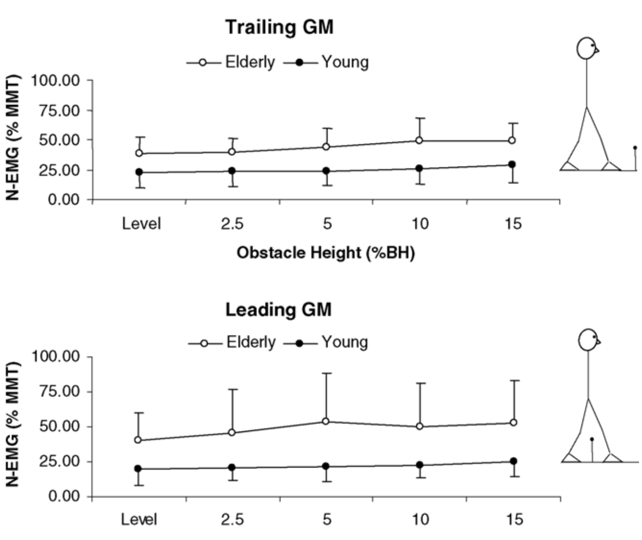

Fig. 2. Normalized EMG activation percentages of the GM during double-support phases of the crossing stride. Elderly adults required significantly greater percentages of their capacity during level walking and obstacle crossing tasks. Activation levels increased linearly as obstacle height increased.

4. Discussion

As elderly adults cross obstacles, they encounter essentially the same mechanical challenges as young adults of similar stature (assuming similar gait velocities). As there were no significant differences in gait velocity between the two groups ($p = 0.182$), it can be assumed that the mechanical challenges encountered by the joints were similar for the young and elderly adults. Considering that the maximum available strength has been shown to be lower in elderly adults [1,2], it follows that the elderly will use a greater percentage of their neuromuscular capacity to successfully ambulate and safely cross over obstacles. Results from this study revealed that there were age-dependent increases in the percentage of muscular capacity used to cross obstacles. Specifically, the gluteus medius and the vastus lateralis were activated to a significantly greater percentage of maximum capacity in the elderly. Furthermore, as obstacle height increased, the relative activation increased for each muscle tested, indicating substantial challenge encountered by the neuromuscular system in maintaining balance during the dynamic task of stepping over obstacles. These findings are supported by joint strength data further compiled from the two subject groups (see Fig. 3). Isometric strength testing on a KinCom dynamometer (Rehab World, Hixson, TN) revealed a significant age-related reduction of strength (normalized to individual body weight) during HA, KE and APF (one-tailed t-test; $p < 0.001$, 0.001, and 0.002, respectively).

It was unexpected that walking speed would reveal no significant group differences. However, walking speeds of older adults in the present study compare favorably with those reported by McFadyen and Prince [22], and the young adult group demonstrated speeds comparable to those reported by Chen et al. [15]. Inter-laboratory variation might

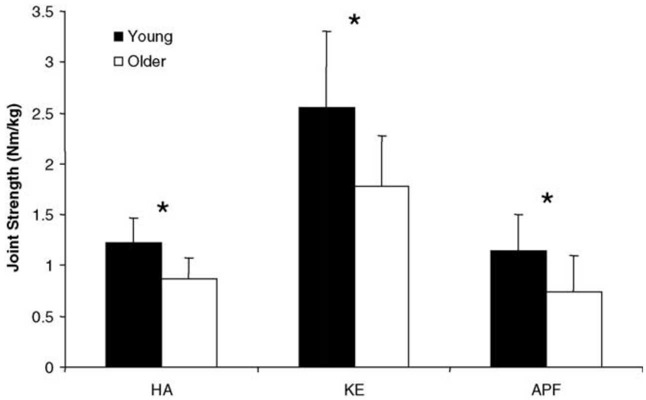

Fig. 3. A comparison of isometric maximal joint strengths between the young and older adult groups (normalized to body weight). Older adults demonstrated significantly lower strength values for each of the joint motions tested ($^{*}p \leq 0.002$).

be used to explain the relative equalizing of gait velocities between the two age groups. Another explanation for the non-significant velocity differences might be the relatively vigorous activity level of our older adult sample. The fact that their gait velocities were not significantly lower than younger adults may indicate that they represent a more able-bodied section of the broader elderly population. It is quite possible that a sample of less active elderly individuals would demonstrate significantly slower walking speeds.

Age-related increases in relative activation of the gluteus medius occurred during the weight acceptance and transfer phase of gait, indicating the critical role played by hip abductors in maintaining medio-lateral (M-L) stability as the center of mass shifts rapidly between each foot. The timing of EMG activation peaks was in agreement with previous work [23,24]. Magnitudes of our N-EMG data were also in agreement with Dubo and colleagues' [24] values for the VL (~25% and 35% for young and elderly adults, respectively), however, our GA magnitudes (~36% and 45% for young and elderly adults, respectively) were noticeably lower than Dubo's (63%). As we measured activation of the medial head and Dubo measured activation of the lateral head (or triceps surae), there may be some inherent discrepancy in the two sets of data. Further comparisons with Dubo's results are difficult to make because of the broad age range of their sample (8–72 years).

It was recently reported that elderly adults with balance disorders displayed greater ranges of motion and higher velocities in M-L motion of the center of mass while stepping over obstacles [20]. Increases in M-L COM motion would certainly increase the demand on hip abductors during weight transfer. Recent results showed that healthy elderly adults demonstrate slightly greater displacements and velocities in the frontal plane (as compared to young adults), however, these differences were not significant [16]. This indicates that while healthy elderly adults may not allow critically greater M-L COM motion, they may be showing the effects of increased challenge in the task of balance control, indicated by higher demand on the neuromuscular capacity of the hip abductors.

One potential limitation of this study is the variable nature of MMT collection. Although the same examiner performed MMT on each subject, no simultaneous measurement was made of the force applied by the examiner, removing the possibility for quantifying variation during MMT testing. However, it was noted that EMG signals consistently leveled off during maximal contractions, indicating that maximum effort was being made by the subjects. A second potential limitation exists in elderly subjects having more adipose tissue over the GM muscle tissue. Care was taken to accurately place the GM electrode and its ability to pick up GM muscle activation was rigorously checked prior to MMT and dynamic trials. Normalizing EMG signals from adipose-rich tissue should be similar to signals from other tissue, as sub-maximal signal magnitudes are simply divided into the signal maximum. The strength of normalization is the ability to report data with respect to present condition.

The N-EMG data presented in this study may be used to quantify the level of challenge on the neuromuscular system when interpreted in light of available joint strength. Although surface EMG has been known to exhibit high variability, it appears that with careful normalization to maximal voluntary activation levels, these proportional values provide a reasonable representation of the control system's response to the challenge posed by daily activities. Furthermore, the magnitude of demand/capacity ratio presented here appears more reasonable than joint moment percentages reported by Bus et al. [25]. Their results indicated that several activities of daily living required greater than 100% of the available torque production in a joint (per maximal voluntary contraction, using a dynamometer). The reason for these high values may arise from the necessary assumptions made in joint torque measurement and estimation techniques [25]. Further study is certainly necessary to resolve discrepancies between measured dynamometric torque and estimated joint moments. Additionally, comparison between these studies suggests that further research is necessary to compare N-EMG values calculated using MMT with those which may be calculated from normalizing to maximal dynamometric contractions. As EMG is a direct measure of muscle activation, perhaps its use is more reliable for representing musculo-skeletal challenge than methods relying on dynamometric joint torque and joint moment estimation techniques, which require many assumptions to be met [26].

The results of this study indicate that healthy elderly adults do require greater percentages of their neuromuscular capacity during level walking and obstacle crossing tasks than young adults. Specifically, the gluteus medius was significantly challenged when the older adults stepped over obstacles. Bus et al. [25] reported significantly greater hip abductor moments in active elderly adults during stair ascent, stair descent and level walking, compared to young adults. Our results support their suggestion that hip abductors should be considered in the functional assessment of elderly adults, and targeted as a critical component of resistance strength training in the older adult population.

As stepping over a higher obstacle poses a greater challenge to dynamic balance control, it may be inferred that decreased lower extremity muscle strength places the elderly population at greater risk for falls. These findings are in agreement with recent work by Lamoureux et al. [8], showing strong association between lower extremity isometric strength and elderly individuals' ability to negotiate obstacles. Combined with the additional findings of Lamoureux et al. [9], the present results add emphasis to the need for muscle strengthening as a preventative intervention, thereby providing improved function in the ambulatory tasks of daily living.

In conclusion, the N-EMG data reported in this study were used to quantify the level of muscular challenge imposed on an individual in relation to their strength capacity. While healthy elderly adults were able to successfully negotiate obstacles of different heights during walking, their muscular strength capacity was significantly lower than young adults, resulting in higher relative muscular demands. Higher demands placed on lower extremity muscles while stepping over obstacles may increase the likelihood of muscle fatigue during daily ambulation, thus placing elderly individuals at higher risk for trips and/or falls. These findings provide further justification for including lower extremity resistance training in the list of preventative interventions for the elderly community.

Acknowledgements

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