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The influence of mechanical vibration on local and central balance control

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ABSTRACT

Fall prevention has an indispensable role in enhancing life expectancy and quality of life among older adults. The first step to prevent falls is to devise reliable methods to identify individuals at high fall risk. The purpose of the current study was to assess alterations in local postural muscle and central sensory balance control mechanisms due to low-frequency externally applied vibration among elders at high fall risk, in comparison with healthy controls, as a potential tool for assessing fall risk. Three groups of participants were recruited: healthy young (n = 10; age = 23 ± 2 years), healthy elders (n = 10; age = 73 ± 3 years), and elders at high fall risk (n = 10; age = 84 ± 9 years). Eyes-open and eyes-closed upright standing balance performance was measured with no vibration, 30 Hz, and 40 Hz vibration of Gastrocnemius muscles. When vibratory stimulation was applied, changes in local-control performance manifested significant differences among the groups ($p < 0.01$). On average between conditions, we observed 97% and 92% less change among high fall risk participants when compared to healthy young and older adults, respectively. On the other hand, vibration-induced changes in the central-control performance were not significant between groups ($p \geq 0.19$). Results suggest that local-control deficits are responsible for balance behavior alterations among elders at high fall risk and healthy individuals. This observation may be attributable to deterioration of short-latency reflexive loop in elders at high fall risk. On the other hand, we could not ascribe the balance alterations to problems related to central nervous system performance or long-latency responses.

1. Introduction

Fall prevention has an indispensable impact on enhancing life expectancy and quality among older adults (Liu-Ambrose et al., 2015; Mohler et al., 2016). One of the early steps to prevent falls is to devise reliable methods to identify individuals at high fall risk, and the underlying mechanisms, thereby supporting targeted intervention. Balance disturbance using mechanical vibration within upright standing tests may provide insights regarding sensory deficits among individuals at high fall risk.

High-frequency (over 60 Hz), low-amplitude (around 1 mm) vibratory stimulation of muscles may lead to kinesthetic illusion (Feldman and Latash, 1982; Goodwin et al., 1972; Latash and Zatsiorsky, 2016; Mancheva et al., 2017; Roll and Vedel, 1982). Herein, the central nervous system (CNS), which is being bombarded by action potentials from muscle spindle endings, may misconstrue that muscle lengthening has occurred. As a result, deceptive joint configurations may be perceived (Craske, 1977). When vibratory stimulation is applied to a postural muscle (e.g., Gastrocnemius, or Tibialis anterior), significant adjustment in the activation of other postural muscles may occur (Hayashi et al., 1981), as well as an increase in center of gravity (COG) sway (Capicikova et al., 2006; Naka et al., 2015; Polonyova and Hlavacka, 2001; Wierzbicka et al., 1998).

An increase in body sway due to low-frequency (less than 20 Hz) vibratory stimulation has also been reported (Naka et al., 2015). Since in this case the sensory signals received by the CNS are not significantly beyond the physiological range, it is not clear whether this increase can be attributed to kinesthetic illusion or to the role of local muscle control. To more fully examine this effect, a

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comprehensive analytic method that incorporates these mechanisms within posturography is essential.

One method to describe balance mechanisms is to divide postural balance control during upright standing into both local postural muscle control (local-control), and higher central nervous sensory feedback cueing (central-control) (Collins and De Luca, 1993; Newell et al., 1997; Toosizadeh et al., 2015b). The local-control stage theorized to work without recruiting sensory feedback from visual, vestibular, and/or somatosensory systems (Collins and De Luca, 1993; Priplata et al., 2002; Toosizadeh et al., 2015b). This mechanism is assumed to operate by setting an activity level required for postural muscles to control short-term body fluctuations (Collins and De Luca, 1993). On the other hand, the central-control mechanism may be called into play in longer time-intervals of body sway, drawing upon sensory feedback to control balance (Collins and De Luca, 1993, 1995; Priplata et al., 2002). To investigate the quality of balance control, separately, within the local- and central-control stages, stochastic stabilogram diffusion analysis has been utilized (Collins and De Luca, 1993). In this method, the temporal displacement between adjacent positions of the COG is used to generate a stabilogram diffusion plot. Next, employing a bilinear mathematical model, local postural muscle control (short time-interval COG displacements) and central sensory feedback balance control (long time-interval COG displacements) paradigms are identified (Collins and De Luca, 1993; Priplata et al., 2002; Toosizadeh et al., 2015b).

Both local- and central-control could be engaged during vibratory stimulation of postural muscles (Latash and Zatsiorsky, 2016; Mancheva et al., 2017; Naka et al., 2015); therefore, when equipped with the aforementioned analysis, vibratory stimulation could be employed to evaluate the quality of different mechanisms responsible for balance. Previous work suggested that among healthy young individuals, mechanical vibration can influence both short-term local muscle control by altering sensory units (Naka et al., 2015), as well as by causing illusionary cueing for central-control (Latash and Zatsiorsky, 2016; Mancheva et al., 2017). However, the balance alterations due to low-frequency vibratory stimulation from a local- and central-control perspective, is still unclear. Further, no study has investigated the effect of vibration stimulation on postural balance among high fall risk older adults, in comparison with healthy individuals. The purpose of this study is to examine the effect of low-frequency, low-amplitude vibratory stimulation of both Gastrocnemius muscles among three groups: (1) healthy young, (2) healthy older adults, and (3) older adults at high fall risk. Through this analysis, balance alterations are analyzed within both local- and central control paradigms.

2. methods

2.1. Participants

Three groups of participants were recruited: healthy young adults (18–30 years), healthy older adults (≥65 years), and high fall risk older adults (≥65 years). Sequentially enrolled subjects underwent blocked group assignment based upon the Center for Disease Control and Prevention's STEADI Risk for Falling Assessment (Rubenstein et al., 2011). Each participant was asked four questions, and one point was assigned for each affirmative answer. The questions were as follow: (1) Have you fallen in the past year? (2) Have you had two or more falls? (3) Are you worried about falling? (4) Do you feel unsteady when you are walking? Participants with a score of zero or one without a history of falling were considered low fall risk, and those with a score of two to four were considered high fall risk. Disorders associated with severe motor deficits and balance performance, including stroke, Parkinson's disease, severe arthritis in lower-extremities, diabetic neuropathy, vestibular diseases, and lower-extremity ulceration and amputation were regarded as exclusion criteria for all three groups. Also, an additional exclusion criterion of an incident fall in the prior year was applied to both healthy young and older adult controls. All participants were recruited after completing written informed consent according to the principles expressed in the Declaration of Helsinki (Association, 2013), as approved by the University of Arizona's Institutional Review Board.

2.2. Clinical measurements

Participants were first asked to fill out clinical questionnaires, including: (1) the visual analog pain scale for lower-extremity (VAS-10) (0: no pain – 10: extreme pain) (Langley and Sheppeard, 1985) within the prior two week period and at the time of the visit; (2) the short falls efficacy scale-international (Short FES-I) to assess the fear of falling (Kempen et al., 2008); and (3) the four-question STEADI-based fall scale described above.

2.3. Balance assessments

Next, each participant performed eight 30-s trials of balance assessment, including: two practice trials with no vibration system attached (one eyes-open and one eyes-closed), two trials with vibration system on calves but with no stimulation (one eyes-open and one eyes-closed), two trials with 30 Hz vibration (one eyes-open and one eyes-closed), and two trials with 40 Hz vibration (one eyes-open and one eyes-closed). Of note, data from practice trials were not used in analyses. In each trial, participants stood upright with their arms crossed across their chest while their feet were kept as close together as possible without touching. The center-of-gravity (COG) was estimated using wearable motion sensors following the same procedures reported earlier (Najafi et al., 2010; Toosizadeh et al., 2014; Toosizadeh et al., 2015c). Briefly, two sensors, each including a tri-axial gyroscope, were used to estimate three-dimensional ankle and hip angles (BalanSens™; BioSensics, Boston, USA). A two-link inverted-pendulum model of the human body was used to calculate the COG from anterior-posterior and medial-lateral angles of the ankle and hip (Najafi et al., 2010; Toosizadeh et al., 2015a).

In order to investigate the performance of local- and central-control strategies, the stabilogram diffusion analysis has been employed (Collins and De Luca, 1993, 1995). In this method, a bilinear model is fitted to the square of displacement between successive COG data points versus increasing time intervals. The first line represents the rate of body sway in short time-intervals and embodies the local postural muscle control (i.e., local-control). The second part of the model represents the rate of body sway in long-time intervals using higher central nervous system based on sensory cueing postural control (i.e. central-control). In order to assess balance performance within the long time-intervals, as suggested by previous work (Toosizadeh et al., 2015b), a logarithmic scale was used here for curve-fitting within the central-control region. Three extracted parameters from this model included: local-control$_{\text{slope}}$, central-control$_{\text{slope-log}}$, and local-control$_{\text{time-interval}}$ (See the supplementary material for details of parameter estimations). Local-control$_{\text{slope}}$ represents the rate of body sway and accordingly balance performance within the local-control stage in normal scales; central-control$_{\text{slope-log}}$ represents the rate of body sway and balance performance within the central-control stage; and local-control$_{\text{time-interval}}$ designates the maximum time interval in which the central-control mechanism using sensory cueing is added to the local-control mechanism for maintaining balance. For each balance parameter, percentage change was estimated comparing the stimulation condition with the no-stimulation condition. In addition to these parameters, we have computed COG sway as the product of sway range in the anterior-posterior and medial-lateral directions.

We used Velcro straps to attach the vibrators to the belly of the muscles. Using a focal vibrator, mechanical stimulation of 30 Hz and 40 Hz frequencies and 1 ± 0.002 mm amplitude was applied to both Gastrocnemius muscles. The employed focal vibrator generated mechanical stimulation using eccentric rotating servomotor.

Previous studies showed that the body sway when exposed to calf vibration, especially in high frequencies (60–90 Hz), continuously increases after the onset of stimulation and saturates after ~30 min with the presence of vibratory stimulation (Capicikova et al., 2006; Tjernström et al., 2002). Thus, here for balance trials with vibratory stimulation, participants were exposed to one minute warm-up vibration prior to tests to assure effects of stimulation reach a plateau level. Of note, balance measurement experiments were started immediately after the warm-up period. To minimize the residual effects of vibration on balance behaviors (Capicikova et al., 2006; Wierzbicka et al., 1998), participants had a two-minute rest period between trials. Considering residual effects of mechanical vibration and performing within-subject comparisons with the baseline data, we tried to minimize any potential systematic bias of the data, by performing the baseline balance tests prior to any calf vibration, followed by 30 Hz and 40 Hz vibration tests.

2.4. Statistical analysis

Differences in socio-demographic parameters and subjective questionnaires among three balance groups (healthy young, healthy older adults, and high fall risk) were assessed using one-way analysis of variance (ANOVA) models. To assess differences in balance parameters among three groups, first, the Shapiro-Wilk test was used to examine the normality of the postural parameters obtained from the samples. In case normality was confirmed, multivariable repeated measures ANOVA models were used to assess differences in balance behaviors between three balance groups. In case the departure from normal distribution was observed, the Friedman test was employed. In each model, baseline balance parameters (trials with no stimulation) or percentage change in balance parameters due to vibration (compared to the condition with no stimulation) were considered as dependent variables. The balance group, sex, vibration frequency (within subject variable), and BMI were considered as independent variables. Cohen's effect size was calculated for each test. Also, the interaction effect between balance groups and vibration frequency was assessed. In case a significant difference was observed among the groups, we repeated the analysis without adjusting the level of significance, considering only two groups of healthy and high fall risk older adults as the independent variable. Moreover, in order to assess differences in age, similar statistical models were used, replacing age with the balance group. Further, to understand how changes in total body sway were associated with local or central sensory deficits, separate correlations were calculated between COG sway changes with local-control$_{slope}$ and central-control$_{slope\text{-}log}$ changes due to vibration. Data from both 30 Hz and 40 Hz vibration stimulation were used and linear Pearson correlations (r) were derived. All analyses were done using JMP (Version 11, SAS Institute Inc., Cary, NC), and statistical significance was concluded when $p < 0.05$.

3. Results

3.1. Participants

Demographic information and subjective questionnaires are reported in Table 1. Thirty participants were recruited, ten in each group. Age was significantly different between the elder low fall risk versus high fall risk groups: Age 72.90 (2.81), 83.60 (9.46), ($p \leq 0.0001$). As expected fear of falling (FES-I score) and the fall score was significantly different between older adult groups ($p \leq 0.001$, Table 1).

3.2. Balance behaviors among high fall risk and healthy individuals

Baseline balance results (without stimulation) showed that the local-control$_{slope}$ in the high fall risk group was respectively more than one and two times larger than healthy older and healthy young participants, within both eyes-open and eyes-closed conditions (Table 2 and Fig. 1). However, central-control$_{slope\text{-}log}$ and local-control$_{time\text{-}interval}$ were not significantly different among the three groups in these conditions (Table 2). Also, within the eyes-open condition, the mean COG sway was the same among the healthy young and older adult groups; however, a 45% increase in the COG sway of the high fall risk group was observed compared to the healthy individuals (Table 2). This increase became magnified to 60% in the eyes-closed condition (Table 2).

Alterations in balance behaviors were observed when the vibratory stimulation was applied. Similar to baseline balance behaviors, local-control$_{slope}$ was significantly different among the groups, especially within the eyes-closed condition. Local-control$_{slope}$ percentile changes for high fall risk participants showed less change due to vibratory stimulation in comparison to healthy participants (Fig. 2 and Table 3); on average, 97% and 92% less change among high fall risk participants compared to healthy young and older adults, respectively. Since the vibration-induced local-control performance was significantly different among the groups, the analyses were repeated between the high fall risk and healthy older adults group. Here, the local-control$_{slope}$ was only significantly different among these groups within the eyes-closed condition ($p = 0.06$ for eyes-open and $p < 0.01$ for eyes-closed condition).

Unlike local-control$_{slope}$, central-control$_{slope\text{-}log}$ and local-control$_{time\text{-}interval}$ percentile changes due to vibration were non-significant among the groups, within both eyes-open and eyes-closed conditions. Local-control$_{time\text{-}interval}$ percentile changes for high fall risk participants were positive while the corresponding values for the other two groups were negative, suggesting in case of no visual feedbacks due to stimulation, high fall risk participants needed more time to switch from the local-control to the central-control stage.

Both healthy older adult and high fall risk groups showed less COG sway change compared to the healthy young group; on average 67% and 85% less change among healthy elderly and high fall risk participants compared to healthy young, respectively (Table 3 and Fig. 3). Although this parameter showed statistical significance within the eyes-open condition ($p < 0.01$), it was non-significant in the eyes-closed one ($p = 0.06$). Repeated analysis for the eyes-open condition, considering only the high fall risk and healthy elderly participants showed that percentile change of COG sway was not significantly different among these two groups ($p = 0.29$).

Further, for all above analyses, the interaction between balance group and vibration frequency was non-significant ($p > 0.5$).

Similarly, when age, instead of balance groups, was considered as one of the independent variables, regardless of the visual condition, central-control$_{slope\text{-}log}$ was non-significant ($p > 0.40$, Table 3); for local-control$_{slope}$ the results were statistically significant, especially within the eyes-closed condition ($p < 0.001$).

Significant correlations were observed between COG sway changes and local-control$_{slope}$ changes in both eyes-open ($r = 0.67$) and eyes-closed ($r = 0.68$) conditions ($p < 0.0001$, Fig. 4). Corresponding correlations between COG sway changes and central-control$_{slope\text{-}log}$ changes showed weaker associations for

Table 1
Mean (standard deviation or percentage) values of sociodemographic information and subjective questionnaires. A significant difference between three balance groups is presented with the asterisk symbol.

	Healthy young	Healthy older adults	High fall risk	*p*-value
Number, n (% of total)	10 (33%)	10 (33%)	10 (33%)	–
Male, n (% of the group)	5 (50%)	4 (40%)	3 (30%)	0.57
Age, year (SD)	23.30 (2.26)	72.90 (2.81)	83.60 (9.46)	<0.0001*
Stature, cm (SD)	173.16 (9.66)	165.03 (10.91)	165.62 (11.21)	0.18
Body mass, kg (SD)	70.84 (16.72)	64.71 (8.37)	65.24 (16.39)	0.57
BMI, kg/m^2	23.59 (4.81)	23.75 (2.11)	23.52 (4.08)	0.99
Pain at the moment, 0–10 (SD)	0 (0)	0.20 (0.63)	1.90 (2.69)	0.16
Pain within two weeks, 0–10 (SD)	0.10 (0.32)	0.80 (2.53)	3.50 (3.72)	0.15
Short FES-I, 7–28 (SD)	7.30 (0.64)	8.00 (1.63)	14.90 (3.96)	<0.0001*
Fall score, 0–4 (SD)	0.00 (0)	0.10 (0.32)	3.10 (0.74)	<0.0001*
Number of falls within one year(SD)	0.00 (0)	0.00 (0)	2.88 (4.64)	<0.001*

BMI: body mass index.
FES-I: falls efficacy scale-international.
SD: Standard deviation.
* A significant difference is presented with the asterisk symbol.

Table 2
Differences in baseline balance parameters between three groups of healthy young, healthy older adults, and high fall risk older adults. Results are presented for the condition without vibration. Mean (standard deviation) values are presented. A significant difference is presented with the asterisk symbol.

Eyes-open	Healthy young	Healthy older adults	High fall risk	*p*-value†	Effect size
Local-control$_{slope}$ (cm^2/sec) $\times$ 10^{-2}	0.61 (0.29)	1.01 (0.79)	2.15 (2.16)	0.01*	0.60
Central-control$_{slope\text{-}log}$	0.30 (0.20)	0.24 (0.15)	0.23 (0.12)	0.53	0.19
Local-control$_{time\text{-}interval}$ (sec)	0.95 (0.86)	1.21 (1.00)	1.11 (0.94)	0.82	0.12
COG Sway (cm^2)	0.40 (0.27)	0.40 (0.23)	0.58 (0.43)	0.31	0.27
Eyes-closed	**Healthy young**	**Healthy older adults**	**High fall risk**	***p*-value†**	**Effect Size**
Local-control$_{slope}$ (cm^2/sec) $\times$ 10^{-2}	1.36 (0.82)	1.96 (1.55)	4.04 (2.96)	<0.01*	0.64
Central-control$_{slope\text{-}log}$	0.20 (0.14)	0.19 (0.08)	0.13 (0.07)	0.29	0.11
Local-control$_{time\text{-}interval}$ (sec)	1.68 (0.98)	1.45 (0.71)	1.57 (0.84)	0.74	0.11
COG Sway (cm^2)	0.65 (0.33)	0.54 (0.33)	1.04 (0.59)	0.05*	0.51

* A significant difference is presented with the asterisk symbol.
† Models were adjusted for balance group, sex and body mass index as independent variables.

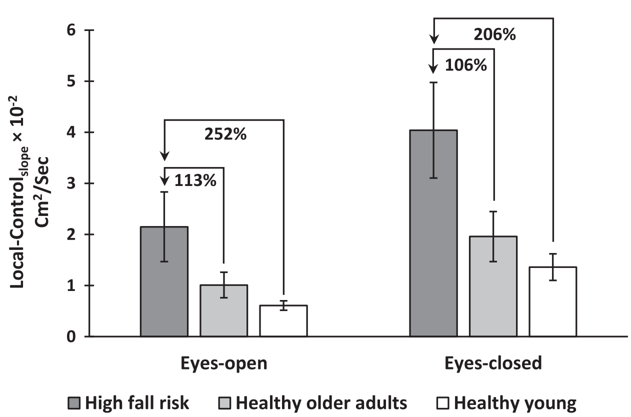

Fig. 1. Mean (standard error) differences in baseline local-control$_{slope}$ among high fall risk and healthy individuals, within eyes-open and eyes-closed conditions.

eyes-open ($r = 0.28$, $p = 0.03$) and eyes-closed conditions ($r = 0.10$, $p = 0.44$).

4. Discussion

Vibratory stimulation of Gastrocnemius muscles influenced high fall risk balance behaviors differently when compared to healthy young and healthy older adults. Results showed that in high fall risk participants when vibratory stimulation was applied to calves, changes in the amount of body sway within short time-intervals or the local-control stage were minimal for the high fall risk group.

Potential reasons for these differences in local-control can be accounted for in the following ways. During upright standing small angular deviations in the ankle joint continuously occur. These deviations increase the length of some lower-extremity muscles. Consequently, spindles of the corresponding muscles become stretched and that enhances the alpha-motoneuron activation, which leads to a muscle force increase (Horak and Nashner, 1986; Matthews, 2011). An opposing torque is required to secure the relatively fixed joint location, which is needed to maintain upright standing; therefore, a co-contraction in antagonistic muscles occurs (Wierzbicka et al., 1998). This reflex activity in older adults is diminished for many reasons including: reduction in number of alpha- and gamma-motoneurons (Kawamura et al., 1977a; Kawamura et al., 1977b), decrease in number of sensory neurons (Maisonobe et al., 1997), and desensitization of muscle spindles (Mynark and Koceja, 2001; Swash and Fox, 1972). These degenerative processes may affect the local-control stage, which may attribute to the higher body sway within the local-control stage in healthy elderly in comparison to healthy young adults. Assuming an exacerbated reflex activity among high fall risk elderly and employing a similar argument, as expected, the highest body sway values during local-control was observed in the high fall risk group.

Another possible explanation for vibration-induced balance alterations among the three groups is related to the central nervous system performance and long-latency responses. Muscle spindles convey sensory information to the central nervous system regarding the level of motor activities (Matthews, 2011). To maintain balance, this information is used in the cerebral cortex to adjust muscle activations (Mihara et al., 2008). As stated in

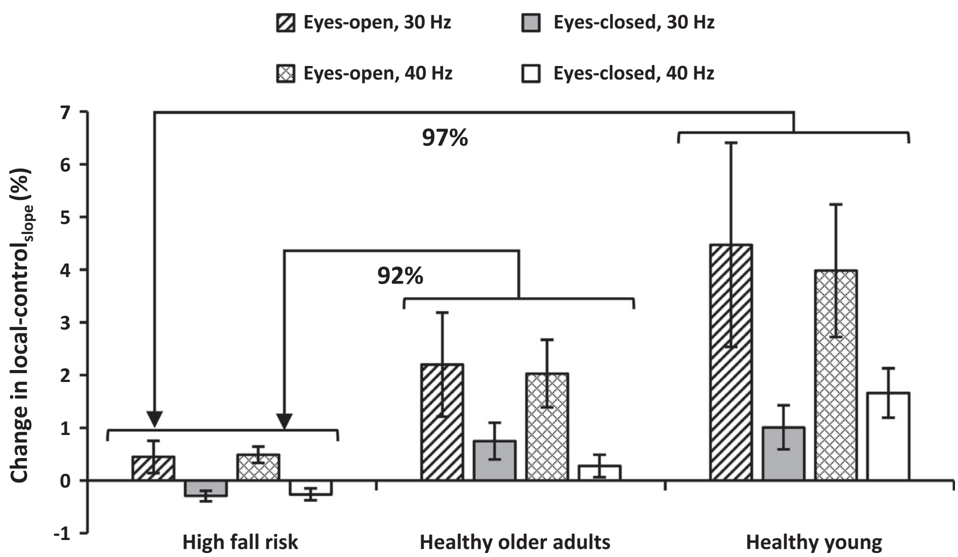

Fig. 2. Local-control$_{slope}$ percentile changes due to vibration. Mean (standard error) differences between three balance groups (healthy young, healthy older adults, and high fall risk older adults) are presented.

Table 3
Differences in balance behaviors between three groups of healthy young, healthy older adults, and high fall risk older adults. Mean (standard deviation) values of balance parameter changes due to vibration are presented. Unless mentioned otherwise, percentile changes of parameters are represented. A significant difference is presented with the asterisk symbol.

Eyes-open	Healthy young	Healthy older adults	High fall risk	*p*-value[†]	Effect size[†]	*p*-value[††]
Local-control$_{slope}$						
30 Hz (%)	4.47 (6.12)	2.20 (3.12)	0.45 (0.97)	<0.01*	0.52	0.01*
40 Hz (%)	3.98 (3.45)	2.03 (2.99)	0.49 (0.97)			
Central-control$_{slope\text{-}log}$						
30 Hz (%)	0.20 (1.75)	0.23 (1.41)	−0.22 (0.99)	0.19	0.18	0.42
40 Hz (%)	0.63 (2.05)	0.43 (1.01)	−0.08 (1.07)			
Local-control$_{time\text{-}interval}$						
30 Hz (%)	1.31 (2.98)	0.98 (1.99)	0.99 (1.86)	0.72	0.05	0.96
40 Hz (%)	0.26 (0.88)	0.84 (2.07)	1.04 (1.73)			
COG Sway						
30 Hz (%)	3.91 (6.53)	0.61 (0.53)	0.59 (1.20)	<0.01*	0.56	<0.01*
40 Hz (%)	1.94 (1.81)	0.67 (0.92)	0.40 (1.03)			
COG Sway (Actual value)						
30 Hz (cm^2)	1.35 (0.99)	0.74 (0.27)	1.15 (1.08)	–	–	–
40 Hz (cm^2)	1.29 (0.71)	0.92 (0.40)	0.85 (0.60)			
Eyes-closed	**Healthy young**	**Healthy older adults**	**High fall risk**	***p*-value[†]**	**Effect size[†]**	***p*-value[††]**
Local-control$_{slope}$						
30 Hz (%)	1.01 (1.31)	0.75 (1.11)	−0.29 (0.32)	<0.0001*	0.76	<0.001*
40 Hz (%)	1.66 (1.48)	0.28 (0.67)	−0.26 (0.35)			
Central-control$_{slope\text{-}log}$						
30 Hz (%)	0.74 (2.08)	−0.20 (0.88)	−0.19 (0.58)	0.80	0.16	0.44
40 Hz (%)	−0.24 (0.84)	0.12 (0.79)	−0.01 (0.65)			
Local-control$_{time\text{-}interval}$						
30 Hz (%)	−0.45 (0.40)	−0.33 (0.28)	0.26 (1.08)	0.06	0.43	0.01*
40 Hz (%)	−0.02 (0.67)	−0.09 (0.49)	0.43 (0.83)			
COG Sway						
30 Hz (%)	0.93 (1.33)	0.67 (1.68)	−0.05 (0.30)	0.06	0.34	0.06
40 Hz (%)	0.86 (1.00)	0.59 (1.00)	0.24 (0.60)			
COG Sway (Actual value)						
30 Hz (cm^2)	1.41 (0.97)	0.93 (0.64)	1.24 (0.60)	–	–	–
40 Hz (cm^2)	1.52 (1.07)	1.10 (0.52)	1.62 (1.18)			

[*] A significant difference is presented with the asterisk symbol.
[†] Models with balance group, sex, vibration frequency (within subject variable) and body mass index as independent variables.
[††] Models with age, sex, vibration frequency (within subject variable) and body mass index as independent variables.

previous studies, vibratory stimulation of muscles can cause some illusionary sensation in the brain regarding the lower-extremity position (Goble et al., 2009; Roll et al., 1989). Previous studies showed aging-induced alterations in the central nervous system, such as decreased attentional resources and a general loss of neural substrate (Raz and Rodrigue, 2006; Toosizadeh et al., 2016).

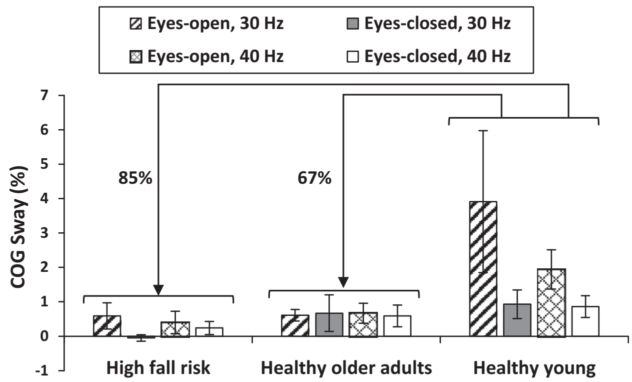

Fig. 3. COG Sway percentile changes due to vibration. Mean (standard error) differences between three balance groups (healthy young, healthy older adults, and high fall risk older adults) are presented.

Therefore, it might be that vibration causes less illusionary disturbance within the central nervous system among elders with impaired balance since messages from spindle units are weaker and the central nervous system may be less sensitive to the disturbance of these messages. However, the results of this study showed no significant differences in the central-control stage among the healthy participants and high fall risk older adults.

Findings here suggest that differences in balance performance between three groups were more detectable within the eyes-closed condition. This is in agreement with previous studies showing that age-related deficits in balance performance and alterations in balance behaviors due to vibration were manifested without visual input (Capicikova et al., 2006; Smetanin et al., 2004).

Similar to the local-control$_{slope}$, the overall sway of the COG showed significant differences between the high fall risk and healthy elderly within the eyes-open condition; however, the performance of the local-control$_{slope}$ to highlight the difference among these two groups is better. Furthermore, while in the eyes-closed condition COG sway was non-significant, the local-control performance manifested a strong difference among the high fall risk and healthy older adults. Besides, comparing only the healthy older adults and high fall risk groups, while COG sway changes were non-significant, local-control$_{slope}$ percentile changes due to vibration showed significant differences within the eyes-closed condition ($p < 0.01$). Therefore, based on the findings of this study local-control$_{slope}$ was shown to be a better-suited parameter to discriminate between high fall risk and healthy elderly while using vibratory stimulation.

As previously reported by Collins et al., the natural aging process affects the operational characteristics of local-control during upright standing (Collins et al., 1995). Here, we did not adjust our first statistical model with age, therefore, we performed a second statistical analysis to examine the effect of age. Although changes in the local-control performance revealed significant differences in this analysis as well, the results were not as strong as the balance group analysis, especially in the eyes-closed condition (Table 3). In other words, we cannot solely attribute the observed differences in the balance group analysis to age effects, alone.

4.1. Limitations and future direction

In this study, high fall risk participants were selected based on their history of falling. We excluded subjects with severe motor deficits, thus we cannot generalize to these groups. In addition, we did not perform a full physical exam to exclude other causes of balance deterioration (peripheral versus central nervous system). Therefore, the current findings require further investigation of clinically confirmed disorders to assure wider validity and generalizability. Another limitation of the current study was the small number of participants in each group, therefore, results should be confirmed in larger samples, and also in comparison with other valid approaches for identifying high fall risk older adults. Further, as explained before, in order to reduce the effect of vibratory residual effects we used non-randomized trial orders which may have served to bias our findings. By performing within-subject comparisons with the baseline, we have tried to minimize confounding error. We recognize that performing the experiments in random order or on different days instead of in one session should be considered as an alternative solution in future studies.

Although we modeled local- and central-control strategies in vibration-induced balance control, we did not perform any direct testing of proprioceptive alterations in response to vibration. Although more direct measurements of sensory deficits are required to confirm our hypothesis, we believe stronger correlations between COG sway changes and local-control$_{slope}$ changes, suggest that most of vibration-induced balance alterations happened due to local-control disturbance. Overall, encouraging though the findings of this study are, they should be interpreted with caution. Specifically, the accuracy of mechanical vibration as a testing tool for assessing high fall risk older adults should be confirmed using direct validated measures of deficits in peripheral (as well as central) nervous system.

The results of this study suggest that the differences in balance behaviors due to low-frequency vibratory stimulation stem from local-control mechanisms and not from central-control ones. Of note, the performance of central-control pathways under stimulation with higher frequency and/or amplitude should be further investigated. Also, employing appropriate mathematical models to represent the musculoskeletal system and its intrinsic

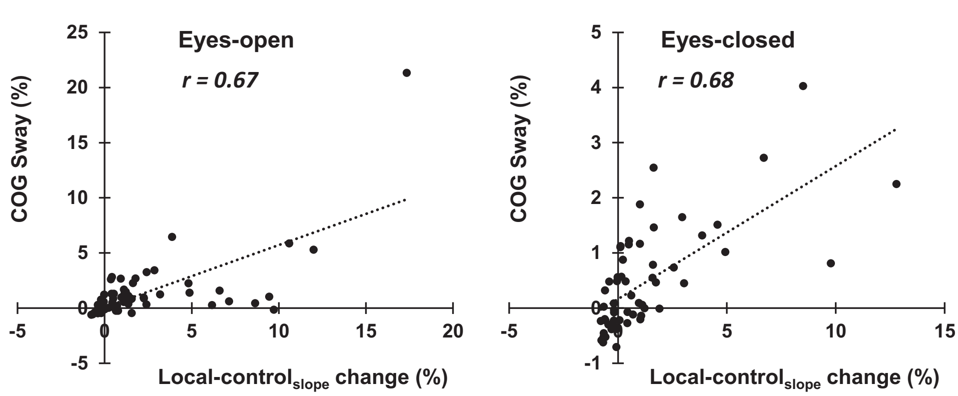

Fig. 4. Correlations between COG sway changes and local-control changes due to vibration stimulation.

properties could be useful to corroborate the significance of local-control mechanism (Ehsani et al., 2016a; Ehsani et al., 2016b; Loeb et al., 1999; Winters, 1995).

4.2. Clinical implications

Our findings are consonant with previous evidence that low-level vibratory stimulation can increase the sensitivity of the human somatosensory system (Gravelle et al., 2002). As shown in the current study mechanical vibratory stimulation should be considered for two fall risk approaches: (1) as a just-in-time fall risk screening tool, assessing both balance performance and sensory deficits for fall risk assessment of older adults; and (2) as a potentially supportive intervention to improve balance while standing and walking. More than half of participants from the high fall risk group showed improvements in balance (reduced overall body sway compared to baseline) when they were exposed to 30 Hz mechanical vibration. On the other hand, less than 10% of healthy young or elderly participants combined showed smaller overall body sway after vibration. These improvements in balance may happen due to activation of muscle proprioceptors as a result of vibration. Associations between mechanical calf vibration, vibration frequency and amplitude, and vibration duration with balance improvements are left to be studied in future research.

5. Conclusions

Within the current study, we examined balance behavior alterations due to vibration among three groups of healthy young, healthy older adults, and high fall risk older adults, within the local- and central control framework. While no significant differences were observed in the central-control paradigm that represents balance control using sensory cueing, vibration-induced differences among the groups were evident within the local-control stage or postural muscle balancing. Unlike healthy participants, the high-fall risk elderly were not influenced under mechanical vibrations and the local-control$_{slope}$ percentile changes were minimal for this group. We attribute this rigidity to diminished sensitivity in the peripheral nervous system in older adults with impaired balance. The results of this study provide a proof of concept for the potential use of mechanical vibratory stimulation as an objective screening tool for fall risk assessment in older adults, and potential as a fall risk intervention for those at high fall risk.

6. Conflict of interest

None declared.

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Appendix A. Supplementary material

Supplementary data associated with this article can be found, in the online version, at https://doi.org/10.1016/j.jbiomech.2018.01.027.

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