Standing Balance and Strength Measurements in Older Adults Living in Residential Care Communities

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Abstract

Background: In the last few decades, research related to balance and mobility in older adults has been conducted in lab-based settings on individuals who are community-dwellers. However, much less information about balance and strength has been collected from residents of long-term care facilities, specifically residential care communities (RCC), even though they are at greater risk of falls than those who live in the community. The aim of this study was the validation of measurement of upright balance using an inexpensive accelerometer and to examine the reliability of strength measurements using a load cell that can be used in any RCC facility.

Subjects: Twenty-nine subjects independently living in residential care communities participated in the study (8M/21F, 87±6 years).

Methods: An accelerometer placed on the back of the subjects at waist level measured body sway in the anterior-posterior (A-P) and medial-lateral (ML) directions. The static standing balance tests consisted of four different standing conditions designed to alter the sensory feedback by having subjects stand for 30s on level and foam surfaces with eyes open and closed and standing in different base of support position. The muscle strength measurements included knee extension and flexion, hip extension and flexion, and ankle plantarflexion and extension. The root-mean-square (RMS) and normalized path length (NPL) for sway in A-P and M-L direction was calculated. Non-parametric statistics were used to test for the effects of test conditions. Concurrent validity was examined using Spearman correlation. Intra-class correlation coefficient was used to examine the trial-to-trial reliability of the strength measurements.

Results: Sway increased significantly as the balance conditions became more difficult due to alteration of sensory feedback ($p < 0.001$) or reducing the base of support ($p < 0.001$). The sway measurement was also able to discriminate between frail and nonfrail people. All strength measurements were highly reliable (ICC = 0.93 to 0.99).

Conclusion: Sway increased as the testing conditions became difficult, providing validation of the accelerometry measurements in the RCC population. The portable load cell provides a feasible, reliable, and inexpensive method for testing lower extremity muscle strength.

Keywords: long-term care, residential care communities, accelerometer, standing balance, and muscle strength

INTRODUCTION

In 2012, approximately 2.1 million people lived in long-term care facilities in the US. Two-thirds lived in nursing homes and one-third resided in residential care communities (RCC).[1] An RCC is defined as a facility that provides room, board with at least two meals a day, and help with personal care such as bathing and dressing or health-related services, such as medication management.[1] Although it has typically been reported that older adults residing in long-term care facilities are at greater risk of falling and sustaining an injury compared to community dwellers.[2], data specifically about fall rates in residents in RCC are lacking. Because individuals living in RCC represent a group that is transitioning to needing greater assistance, we can presume that their fall risk is increasing as well. Consequently, it is important to investigate the fall risk factors in this population.

Lower-extremity muscle weakness and balance impairment are two of the many risk factors that have been associated with mobility limitations and falls in older adults.[2–6] In the last few decades, research related to balance and mobility in older adults has been conducted primarily in lab-based settings with individuals who are community-dwellers. The laboratory-based tests are usually expensive and not portable, so the access to these tests is limited for large group of people. Recent technology advancements have provided inexpensive and portable quantitative tools to assess balance, such as accelerometers and the Wii balance board, in older adults,[7–9] and other populations including: people with Parkinson disease,[10] multiple sclerosis,[11] stroke.[12] To our knowledge no study has applied these technological advancements in RCC environments.

The current gold standard method to measure lower extremity muscle strength is using computerized isokinetic dynamometry.[13] The high cost, low portability, and time-consumption are drawbacks that have limited the application of computerized isokinetic dynamometry in residential care communities. Another method to assess strength in clinical setting is manual muscle testing. Although it is the most frequently used technique to quantify muscle strength and easier to use, it lacks sensitivity and responsiveness, and is subject to a ceiling effect.[14,15] Handheld dynamometers have been commonly used in different settings to objectively quantify muscle strength. Even though handheld dynamometers have good reliability in different populations, it has some important limitations, such as difficulty in stabilizing the subject, and it is influenced by the strength of the examiner especially for large muscles group.[16,17] The concept of using a simple strain-gauge uniaxial load cell device has been proposed before but it has not been used with people who live in residential care communities.[18] A uniaxial load cell device provides an easy and reliable way to overcome the aforementioned drawbacks and quantify muscle strength in different settings outside the clinic.

The purpose of this study is the validation of upright balance and lower extremity muscle strength measurements in older adults living in RCCs using inexpensive and portable devices that can be used in any long-term care facility.

METHODS

Subjects and Setting

Twenty-nine participants (8M/21F, 87±6 years) enrolled in this research study. All subjects were recruited using a convenience sample of older adults living in three RCC facilities, who were already enrolled in research study investigating function of residents in RCC settings. Inclusion criteria were (a) age 65 years or older, (b) living independently in RCC facilities, and (c) cognitively able to provide informed consent. Exclusion criteria included (a) subjects who were not able to ambulate for 1 minute (assistive devices were allowed), (b) any medical or neurological condition that prevented subjects from performing maximal muscle action. The study was approved by the Institutional Review Board (IRB) of the University of Pittsburgh.

Instrumentation

1- Balance Accelerometry

Balance accelerometry was adapted from part of the National Institutes of Health (NIH) toolbox project.[19] The system consists of a dual axis accelerometer (ADXL213AE, Analog Devices, Inc., Norwood, MA) oriented to record mediolateral and anteroposterior acceleration of the body. The acceleration is transmitted via a Bluetooth transmitter to a laptop computer at 50 Hz and with 16-bit accuracy. The system was affixed to the subject's back at the level of the iliac crest using Velcro and a gait belt. A custom written Labview program was used to acquire the data (**Figure1**).

2- Uni-axial load cell for assessment of lower extremity strength

A uni-axial load cell (Measurement Specialties ELPF Series) was used to measure the lower extremity strength. The load cell has a maximum capacity of 2225 N. The load cell was connected to an amplifier that displayed the instantaneous and maximum force exerted on the load cell. The load cell was arranged in series with straps (two cuffs) that fit around the limb on one end and a stable object on the other end. **(Figure 2).**

Procedure

Standing Balance Test:

The protocol consisted of multiple tests. The first test was based on the modified Clinical Test of Sensory Interaction in Balance (mCTSIB,),[20] and had 4 different standing conditions designed to alter the sensory feedback: (1) standing with feet together on firm surface with eyes open (FIRM-EO); (2) standing with feet together on firm surface with eyes closed (FIRM-EC); (3)

standing with feet together on foam surface with eyes open (FOAM-EO); (4) standing with feet together on foam surface with eyes closed (FOAM-EC). The foam surface that was used in the testing consisted of an Airex Balance Pad (Advanced Medical Technology Inc., Watertown, MA), with a thickness was 6 cm. In these 4 conditions subjects stood for 30s, as still as possible, facing a wall with their arms crossed in front of their chest. An experimenter guarded the subject from behind in case a loss of balance occurred. In between each condition, subjects had a seated rest break for 30-60 s. If subjects were not able to complete a trial, they were given one other opportunity to perform.

For the second balance test, subjects performed an instrumented Short Physical Performance Battery (SPPB).[21] First, subjects stood for 10 sec with three different bases of support: feet together, semi-tandem (one foot halfway in front of the other), and tandem (heel of one foot directly in front of the toes of the other foot). Subjects could choose which foot to put in front.

If subjects were unable to complete, they continued to the next condition and the investigator documented that subject wasn't able to do the condition.

Lower Extremity Strength Testing:

Strength measurements included three maximum voluntary isometric contractions (MVIC) for six different muscle groups. All of the testing was done in sitting position; details about the device positions for each muscle group are summarized in **Table 1**. The test was done with the subject's dominant leg, which was identified by asking the subject the side of hand dominance. All the trials were done while the subject sitting on chair. The tone and words of encouragement used by the examiner were standardized. Thirty seconds of rest was provided between trials. The average of the three trials was used in the data analysis. All of the measurements were taken by a physical therapist.

Secondary outcome measures:

In order to examine the concurrent validity, measures from the most recent assessment in the parent study were recorded.

Duke Comorbidity Index

The Duke Comorbidity Index is a simple patient self-report measure that contains 18-item list of chronic conditions. The index has been validated for various subgroups such as community-older adults and stroke population.[22,23]

Grip strength A JAMAR PLUS+digital hand dynamometer (Sammons Preston) was used to measure grip strength. Five trials were performed with the dominant hand.

Gait speed

Self-selected gait speed was measured over 4 meters at usual pace. This measure assesses usual walking speed over short distance. It represents a quick, inexpensive and highly reliable tool to use with older adults.[24] Participants stood with both feet touching the start line and asked to walk at his/her usual speed until he/she passed the cone that was placed at the end of the 4 meters. The average of two trials was used in the data analysis.

Fried's Frailty Index

Fried's phenotype of frailty[25] was used to assess frailty; scoring is based on presence of five frailty indicators: weight loss (self-report of unintentional weight loss of 10 pounds or more in the past year), exhaustion (identified by subject responses to the questions from the Center for Epidemiological Studies depression (CES-D) scale: "I felt that everything I did was an effort" and "I could not get going"), slow gait speed (time to walk 15 feet, adjusted for gender and height), weakness (grip strength, an established cutoff stratified by gender and BMI provided by Fried et al.,), and low physical activity (measured using subject responses to a leisure time questionnaire, the Minnesota Leisure Time Activities Questionnaire).[25] Each indicator is assigned a value of 0 or 1. Subjects were divided into three groups based on total score: Nonfrail = 0 indicators, Prefrail = 1-2 indicators, Frail = 3 or more indicators.[25]

Data Analysis

The balance accelerometry data were visually inspected and sudden, extraneous movements were removed. The first and last five seconds for the mCTSIB conditions were not included in the analysis in order to avoid transition effects. Using a custom written Matlab program, the accelerometry data were lowpass-filtered using a 4th order Butterworth filter with a cutoff frequency of 2 Hz. The Root Mean Square (RMS), and the Normalized Path Length (NPL) of the acceleration were calculated for both the anteroposterior (AP) and mediolateral (ML) directions. The RMS and NPL were calculated as follows:

$$RMS = \sqrt{\frac{\sum_{j=1}^{N-1}(a_J)^2}{N}} \quad (1)$$

$$NPL = \frac{1}{T}\sum_{j=1}^{N-1} \left|a_{j+1} - a_j\right| \quad \text{mG/s (2)}$$

Where *N* is the number of samples, and *t* equals to the time duration, and *a* is the acceleration data at time sample *j*.

Statistical Analysis

SPSS version 21.0 was used for all data analyses. Descriptive statistics were calculated for all participants. For the accelerometry data, a nonparametric statistical analysis was used for all comparisons because of the skewed distribution of scores. The Friedman test was used to examine if significant differences existed in the postural sway among the mCTSIB and SPPB balance tasks. Post hoc analysis with Wilcoxon signed ranks test was conducted with a Bonferroni correction applied. In order to assess the test-retest reliability for the lower extremity muscle strength testing across all three contractions, the intra-class correlation coefficient (ICC) was calculated (Model 3,1).[26] A Spearman rho correlation test was used to examine the relationship between postural sway and muscle strength measurements with gait speed, number of comorbidities, grip strength and frailty index. The Kruskal-Wallis test was performed to determine differences in postural sway and strength across frailty groups.

RESULTS

Participant Characteristics

A total of 29 subjects living in RCCs were assessed for the validity of the balance accelerometry and the muscle strength measurements. The average age of the participants was 87 ± 6 years (range: 72-98 years). Seventy-two percent (21/29) of the participants were female. About 38% (11/29) of participants had two or more falls in the previous year, and 28% (8/29) had only fallen once in the last year **(Table 2).** Reduced grip strength was the most common frailty indicator (72% of subjects), followed by reduced walking speed (45%), low physical activity (41%), exhaustion (21%), and unintentional weight loss (3%). These frailty indicators resulted in 4 subjects classified as Nonfrail, 18 subjects as Prefrail, and 7 subjects as Frail.

Balance Accelerometry

A significant increase in RMS and NPL sway in both the AP and ML directions occurred as the balance conditions became more difficult due to alteration of sensory feedback in the mCTSIB ($p < 0.001$), or reducing the base of support in the SPPB ($p < 0.001$). RMS sway in the AP direction was increased by at least 50% when subjects closed their eyes compared with eyes open, or stood on foam compared with standing on level surface. In the ML direction, RMS sway increased by at least 60% with eyes closed, and by at least 100% when standing on foam compared to level surface **(Figure 3a).** The NPL sway in both directions was increased by at least 50% during eyes closed compared with eyes open, and when standing on compliant surface as compared to standing on the firm surface **(Figure 3b).** During the SPPB, RMS and NPL sway in both AP and ML directions were increased by at least 60% when subjects stood in the tandem position compared with standing with feet together and in semi-tandem, and RMS sway in the ML direction was increased by about 40% during semi-tandem stance compared with feet together **(Figure 4 a, b).**

Validation of balance accelerometry

During eyes open on compliant surface, RMS sway in the AP and ML directions was greater in subjects classified as frail compared with those who were pre-frail (AP p=0.025, ML p=0.007), and compared with subjects who were non-frail (AP p=0.021, ML p=0.043, Figure 5). During tandem stance, RMS and NPL sway in the ML direction was larger in the subjects who were pre-frail compared with non-frail (RMS p=0.006, NPL p=0.005), and only for NPL sway between frail and nonfrail (p=0.029, Figure 5). Also, when subjects stood on firm surface with eyes open, RMS in the AP direction was larger in subjects who were frail compared with prefrail (p=0.010) **(Figure 5)**.

RMS sway in the ML direction during semi-tandem stance was significantly different between fallers and non-fallers (p = 0.004), but no other sway measures were associated with fall history. In Table 3, correlations between balance accelerometry measurements and clinical outcome measures are shown. A greater number of comorbidities was associated with increases in both AP and ML sway during foam EC (Spearman rho = 0.39 AP, 0.41 ML), semi-tandem (Spearman rho =0.57 AP, 0.46 ML) and tandem conditions (Spearman rho =0.47 AP, 0.51 ML). Elevated RMS and NPL sway in ML direction during tandem standing was associated slower gait speed (Spearman rho = -0.65 RMS, -0.42 NPL)**.**

Strength measurements

Descriptive statistics and test-retest reliability data for strength measurements are presented in **Table 4**. Test-retest reliability was calculated from 3 repeated tests and all measurements were highly reliable (ICC = 0.93 to 0.99). The ankle dorsiflexion strength was greater in subjects who were pre-frail compared with frail (p=0.015) **(Figure 5)**. An increase in lower extremity strength performance was associated with increased grip strength except for hip flexion strength (Spearman rho= 0.56 – 0.76) and increased gait speed except for ankle plantarflexion and hip flexion (Spearman rho = 0.49 – 0.52), and a decrease in the number of comorbidities was associated with greater knee flexion and ankle dorsiflexion strength (Spearman rho = -0.47, Spearman rho =-0.48) **(Table 5).**

DISCUSSION

The purpose of the present study was to assess the concurrent validity of balance performance using low-cost accelerometers and to test the reliability and concurrent validity of lower extremity muscle strength using load cell, in older adults living in residential care communities. Body sway measured with the accelerometer increased as the balance task became progressively more difficult by reducing the degree of sensory input or by narrowing the base of support. These findings demonstrate the ability of the accelerometer measurements to vary between sensory integration and base of support conditions. In addition, the test-retest reliability for the lower extremity measurements was high across the three trials.

When somatosensory cues relative to a stable surface were reduced by standing on a foam pad, the older adults produced higher body sway than standing on firm surface.[27] The current findings are in accordance with other research using similar technology.[7,8,28,29] This research extends this previous work by demonstrating feasibility in a residential environment and establishing validity in older adults with a mean age of 87. Moreover, our results showed that the NPL for AP sway during the eyes closed on foam condition was greater than the sway of healthy older adults aged 66-85 in a previous study that used the same accelerometer, which further validates the measurements.[30] In order to further establish concurrent validity, the magnitude of postural sway in both AP and ML directions during standing on foam EC, semi-tandem and tandem was correlated with the total number of comorbid medical conditions, indicating that less sway is related to a lower number of comorbidities.

The current findings suggest that the sway measurements are able to distinguish between older adults at different stages of frailty. Postural sway, in both AP and ML directions, was significantly different between frail vs. nonfrail, and prefrail vs. frail. The results should be interpreted cautiously given the small sample size in this study. Nonetheless, these results are in a line with previous studies in terms of differences in postural balance between the non-frail and frail groups, but disagree with regard to differences between the pre-frail and frail people.[31–33] This may be due to the higher number of subjects in the pre-frail group compared to the frail group, as well as different sway measures that have been used in the previous studies.[31–33] In addition, postural sway in ML direction during semi-tandem stance was significantly different between fallers and non-fallers. Semi-tandem stance places more emphasis on the control of stance in the ML direction, which has been shown to be related to fall history in previous studies.[34–36] These results also help to establish the known-groups validity of the sway measurements.

Different statistical techniques to assess the reliability between repeated measurements have been discussed in the literature.[37] The intraclass correlation coefficient (ICC) analysis was specifically developed to examine relative reliability. Therefore, our results indicate excellent relative reliability across the test trials for all the included muscle groups. In addition, the mean force production was consistent with reference values from adults 70-79 years old that used a hand-held dynamometer.[38] Four out of the six strength values (except hip flexion and ankle plantarflexion) were correlated with grip strength and gait speed. It is possible that hip flexion and ankle plantarflexion was not as highly correlated with grip strength because the examiner provided some of the resistance, rather than the cuff being attached to an immovable object. As a result, the examiner may have influenced these recordings.

The gold standard method to objectively quantify postural stability is the use of force platforms. Although, force platforms are a reliable and sensitive tool,[39] its cost, required space, and lack of portability, result in limited clinical usage, as well as limit the ability to use it across different populations such as people who live in residential care communities. An alternative way to overcome these limitations is by using body-warn accelerometers such as the one we used in this study. We were able to go to different facilities and test subjects at their home. Therefore, implementing such technology will help researchers to access understudied populations such as residents of RCC, and allow clinicians to take measurements in real-life environments, and eventually help clinician and therapist to develop a proper intervention based on the subject's balance assessment. Finally, the uniaxial load cell provides a reliable and inexpensive method to quantify muscle strength in different environments outside the clinic.

One issue regarding clinical research in long term care facilities (including RCC) is the range of function of the residents. Long term care centers can include skilled, assisted living, independent, dementia units and short term care. Furthermore residents often transition from 1 level to another depending on acute illness, fractures, family requests or physician input/supervision. Therefore, rather than classify or characterize residents by level of care, functional status or frailty measures are often used to characterize a cohort. Given the important relationships between falls, balance and lower extremity strength, developing low-cost and portable assessments of balance and strength is essential for monitoring the health status of the residents.

A main limitation the current study is the relatively small sample size. For this reason it was difficult to find relationship between some of the outcome measures, for instance, between sway measurements and gait speed in most of the balance conditions. Future studies with larger sample sizes may help to discover more relationships between balance and mobility measurements.

Conclusion

This study showed sway increased as the testing conditions became difficult, and correlated with grip strength, gait speed, and number of comorbidities, providing concurrent validity to the accelerometry measurements in the RCC population. Additionally, the sway was different among the frailty subgroups. Balance accelerometry may serve as a useful biomarker for future mobility problems. In addition, the portable load cell provides a feasible, reliable, and inexpensive method for testing lower extremity muscle strength.

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Tables and Figures

Table 1: Testing positions for strength measurements. Subject was seated in a standard height 4-legged chair in all positions.

Isometric Action	Testing Device Position
Knee extension	The subject was seated with hips and knees flexed 90 deg. One cuff of the device was placed around the chair leg and the other cuff around the ankle of the subject's dominant leg. The subject extended the knee gradually until the maximum force was attained over the course of five seconds
Knee flexion	The subject was seated with hips and knees flexed 90 deg. One cuff was placed around subject's ankle and the other cuff was attached to fixed object in front of the subject. The subject flexed the knee gradually until the maximum force was attained
Hip abduction	The subject was seated with hips and knees flexed 90 deg, with knees touching at midline. One cuff was placed around the right thigh and the other cuff around the left thigh, just proximal to the knee joint. The subject moved both knees laterally until the maximum force was attained
Hip flexion	The subject was seated with hips and knees flexed 90 deg. One cuff was placed around the thigh just proximal to the knee joint and the second cuff was attached to the ground by having the examiner step on it. The subject flexed the hip in a superior direction until the maximum force was attained
Ankle plantarflexion and dorsiflexion	The subject was seated with hips flexed 90 deg. The dominant leg was extended at the knee. One cuff was placed around the distal part of the foot approximately at the first metatarso-phalangeal joint, and the other cuff was attached to a fixed object (e.g. arm of the chair). For ankle plantarflexion, the examiner held a bar with the other cuff attached to it. Subject plantarflexed or dorsiflexed the ankle until the maximum force was attained

Table 2: Demographic and clinical characteristics of subjects

Variable	Mean (SD)	Range
Age (y)	87 (6)	72 - 98
Body mass index (kg/m^2)	28.1 (5.6)	21 - 47
Number of comorbidities	7 (3)	2 - 13

Falls in the last year	Number of subjects
0	10
1	8
2 or more	11

Table 3: Spearman correlation coefficients between balance accelerometry conditions, and Duke Comorbidity Index and gait speed.

	Duke Comorbidity Index	**Gait speed**
Foam EC (AP-RMS)	**0.39**	-0.06
Foam EC (ML-RMS)	**0.41**	0.10
Feet tandem (AP-RMS)	**0.47**	-0.39
Feet tandem (AP-NPL)	**0.57**	-0.38
Feet tandem (ML-RMS)	**0.52**	**- 0.65**
Feet tandem (ML-NPL)	**0.42**	**- 0.42**
Semi-tandem (AP-NPL)	**0.57**	-0.13
Semi-tandem (ML-RMS)	**0.46**	-0.07
Semi-tandem (ML-NPL)	**0.50**	0.08

Bolded numbers indicate significant correlation coefficient $p < 0.05$.

Table 4: Mean (SD) lower extremity strength performance and test-retest reliability, indicated by the intraclass correlation coefficient (ICC) and standard error of measurement (SEM).

Isometric Action (Kg)	Mean (SD)	ICC (CI 95%)	SEM
Knee Extension	16.7 (7.6)	0.95 (0.91-0.97)	1.69
Knee Flexion	12.5 (4.6)	0.98 (0.97-0.99)	0.65
Hip Abduction	18.1 (7.6)	0.99 (0.98-0.99)	2.40
Hip Flexion	14.4 (4.9)	0.97 (0.95-0.98)	0.84
Ankle PF	27.3 (11.7)	0.97 (0.95-0.98)	2.02
Ankle DF	11.8 (4.8)	0.93 (0.87-0.96)	1.26

Table 5: Spearman correlation coefficients between lower extremity strength, and Duke Comorbidity Index, gait speed, and grip strength.

Isometric Action (Kg)	Duke Comorbidity Index	Gait speed	Grip strength
Knee Extension	-0.13	**0.49**	**0.56**
Knee Flexion	**- 0.47**	**0.50**	**0.76**
Hip Abduction	-0.25	**0.52**	**0.68**
Hip Flexion	-0.11	0.23	0.34
Ankle PF	-0.07	0.19	**0.66**
Ankle DF	**- 0.48**	**0.52**	**0.65**

Bolded numbers indicate significant correlation coefficient $p < 0.05$

Figure 1: An accelerometer placed on back of subject, using Labview software to acquire balance data

Figure 2: A uni-axial load cell attached to subject's leg to measure hip abduction muscle strength.

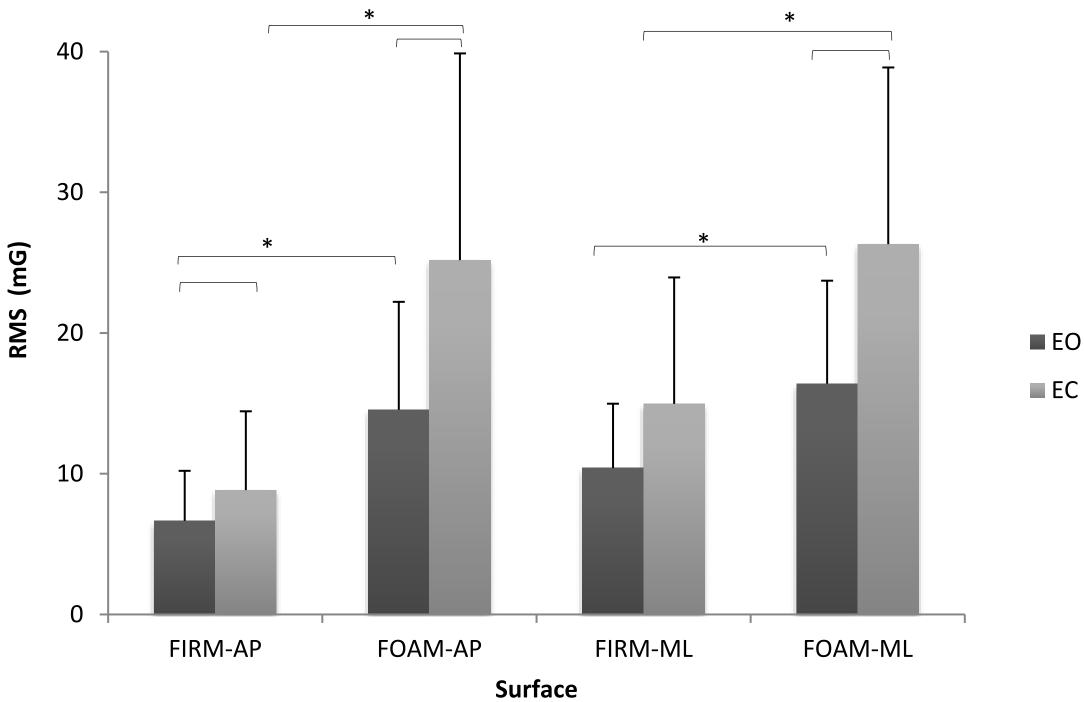


Figure 3a: Effect of vision (Eyes Open: EO, and Eyes Closed, EC) and surface conditions (Firm, Foam) on root-mean-square (RMS) sway acceleration for antero-posterior (AP) and mediolateral (ML) directions. (Error bars represent ± 1 standard deviation).

*: Significant differences with $p < 0.05$.

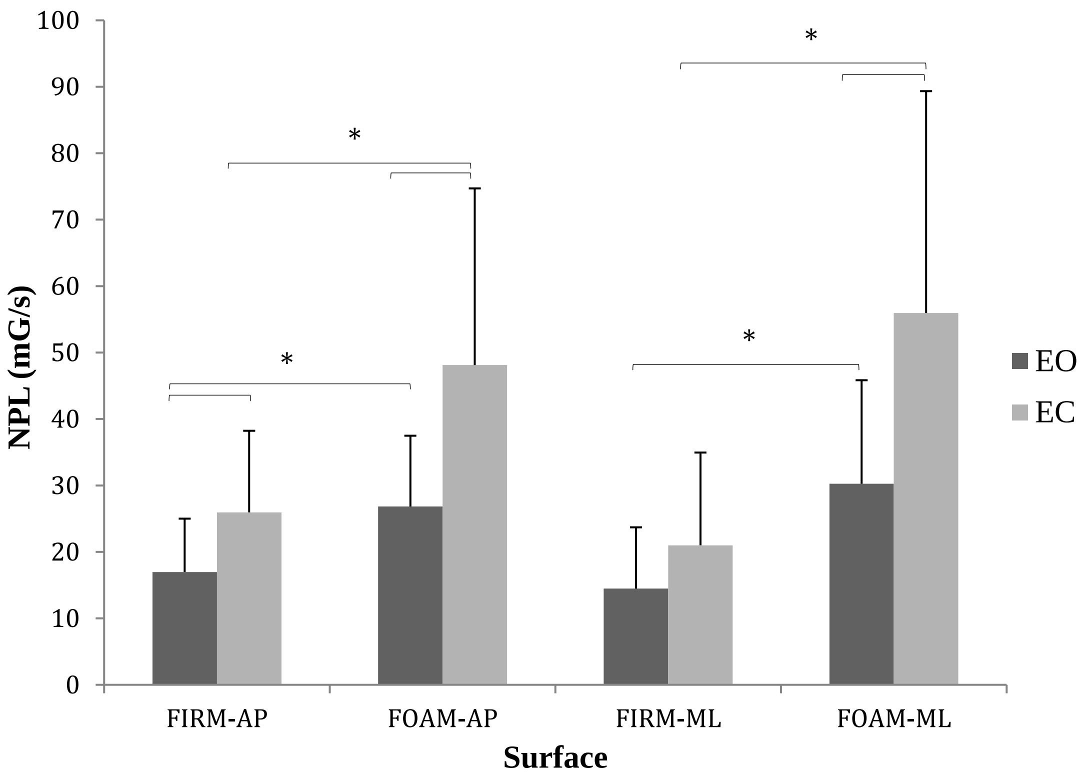


Figure 3b: Effect of vision (Eyes Open: EO, and Eyes Closed, EC) and surface conditions (Firm, Foam) on normalized path length (NPL) sway acceleration for antero-posterior (AP) and mediolateral (ML) directions. (Error bars represent ± 1 standard deviation).

*: significant differences with $p < 0.05$.

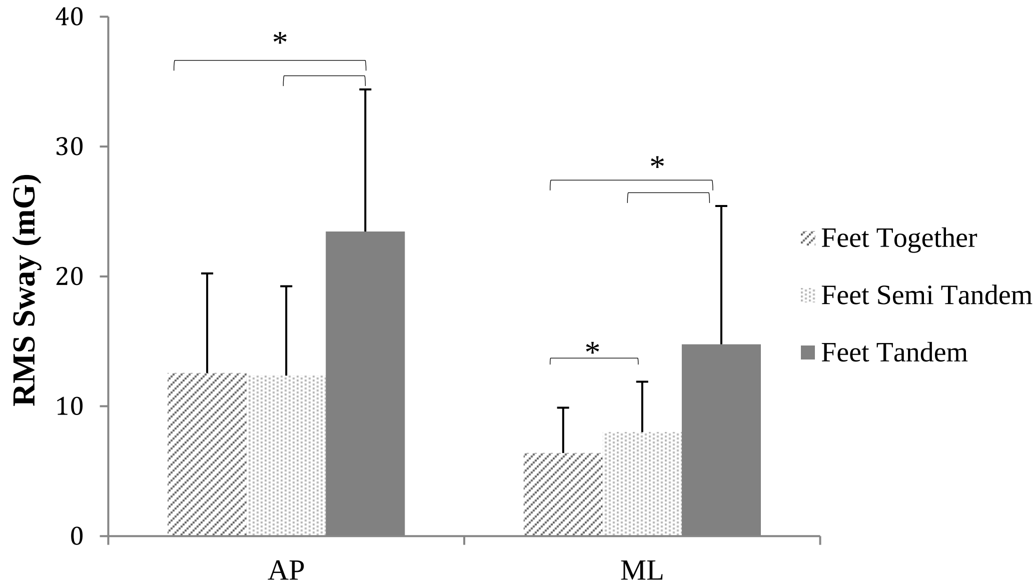


Figure 4a: Effect of base of support on root-mean-square (RMS) sway acceleration for antero-posterior (AP) and mediolateral (ML) directions. (error bars represent ± 1 standard deviation).

*: significant differences with $p < 0.05$.

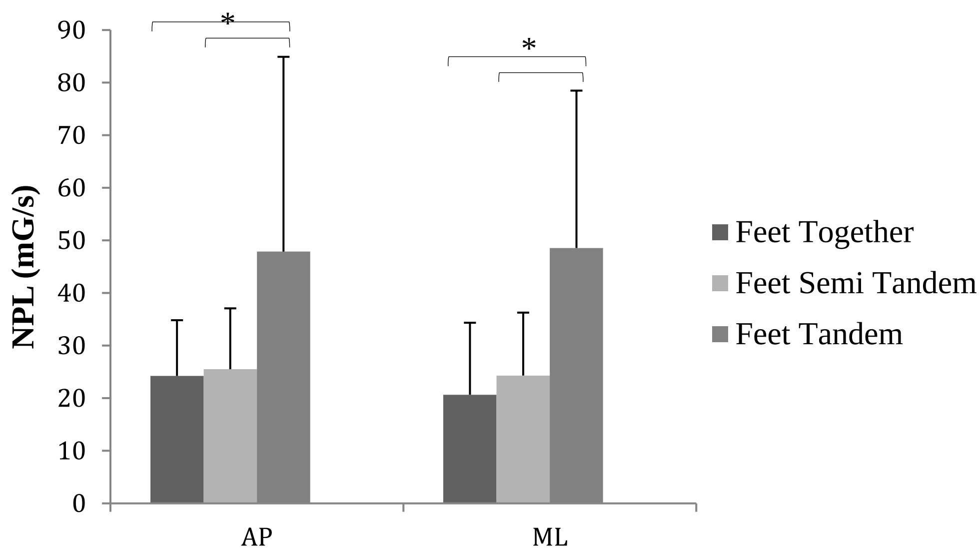
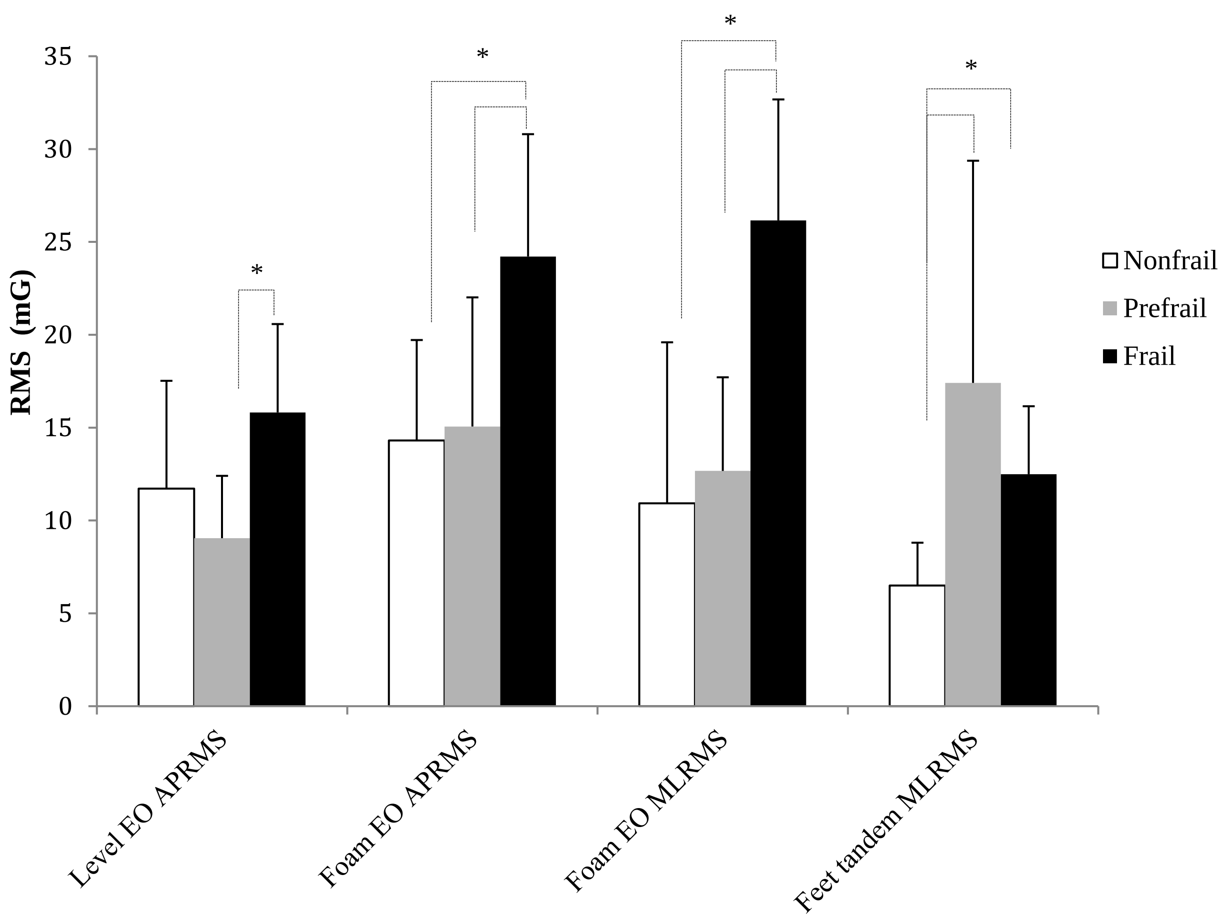


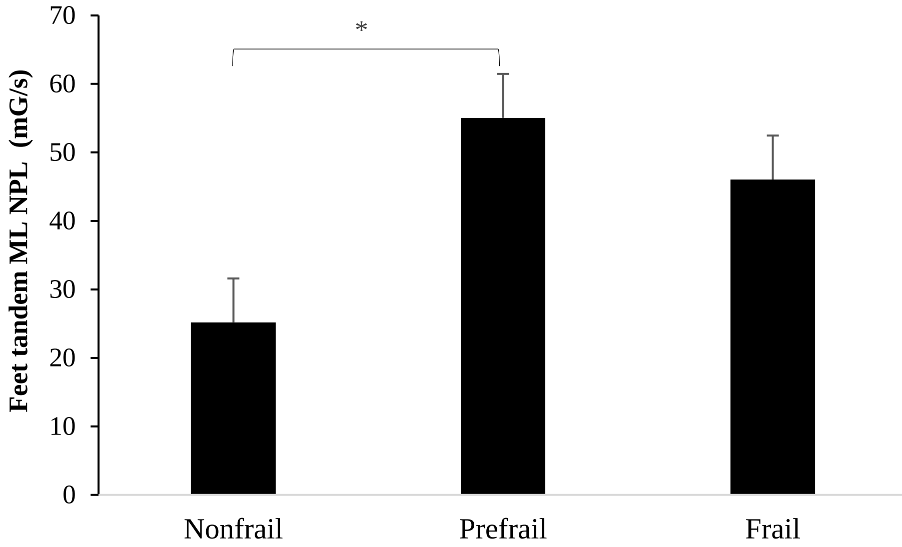
Figure 4b: Effect of base of support on normalized path length (NPL) sway acceleration for antero-posterior (AP) and mediolateral (ML) directions. (error bars represent ± 1 standard deviation).

*: significant differences with $p < 0.05$.

(A)


35
30
25
20
15
10
5
0
RMS (mG)
*
*
*
*
Nonfrail
Prefrail
Frail
Level EO APRMS
Foam EO APRMS
Foam EO MLRMS
Feet tandem MLRMS

(B)



(C)

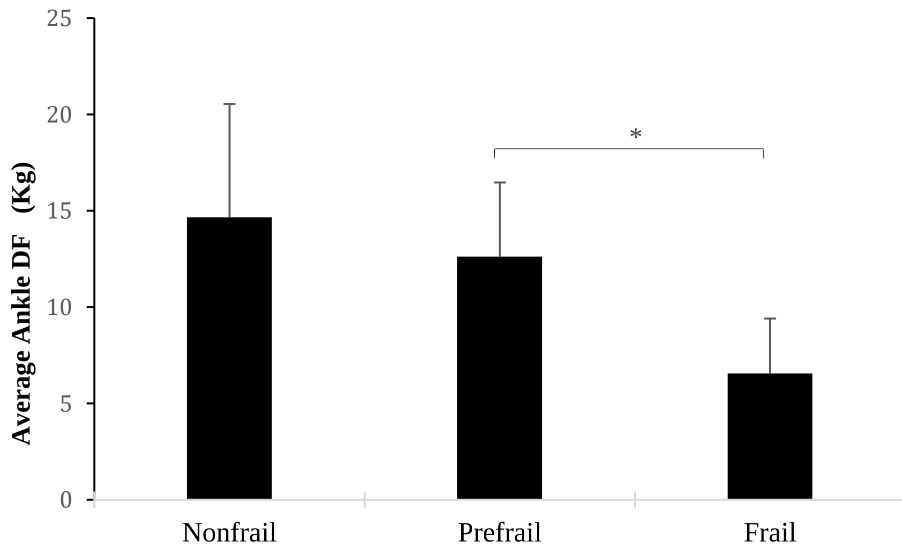


Figure 5: A) Root-mean-square (RMS, mG) sway acceleration during standing on Foam with eyes open (EO), and Level with EO conditions in AP direction and during standing on tandem stance in ML direction across Fried's phenotype of frailty. B) Normalized-path-length (NPL, mG/s) for ML direction during tandem stance across Fried's phenotype of frailty. C) Average ankle dorsiflexion strength (Kg) across Fried's phenotype of frailty

*: significant differences with $p < 0.05$.