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Research paper

Haptic information provided by the "anchor system" reduces trunk sway acceleration in the frontal plane during tandem walking in older adults

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HIGHLIGHTS

- We studied the contribution of an "anchor system" to walking stability in older adults.
- The use of the anchor system reduced trunk sway acceleration.
- This effect did not persist after removal of the anchor system.

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ABSTRACT

This study assessed whether the use of an "anchor system" benefited older adults who performed a tandem walking task. Additionally, we tested the effects of practice with the anchor system during walking on trunk stability, in the frontal plane, of older adults. Forty-four older adults were randomly assigned to three groups: control group, 0 g anchor group, and 125 g anchor group. Individuals in each group performed a tandem walking task on the GaitRite system with an accelerometer placed on the cervical region. The participants in the 125 g anchor group held, in each hand, a flexible cable with a light mass attached at the end of the cable, which rested on the ground. While the participants walked, they pulled on the cables just enough to keep them taut as the masses slid over the ground. The 0 g anchor group held an anchor tool without any mass attached to the end portion. The results of this study demonstrated that the use of the anchor system contributed to the reduction of trunk acceleration in the frontal plane. However, this effect did not persist after removal of the anchors, which suggests that the amount of practice with this tool was insufficient to generate any lasting effect, or that the task was not sufficiently challenging, or both.

1. Introduction

The anchor system, conceptualized by Mauerberg-deCastro [1], is a simple and effective tool used to provide additional haptic information to the postural control system. The anchor system requires an individual to hold the free ends of two flexible cables – one in each hand – which have a light mass (of varying loads) attached to each of the other ends, and which rest on the ground at the individual's sides. During walking and balancing tasks, the individual keeps the mass in contact with the ground while pulling on the cable just enough to keep it taut. The anchor tool has been found to help to reduce postural sway in such tasks. In these anchored balance tasks, body sway causes changes in the cables' tension, and each hand senses the tension through skin receptors. In combination with continuous pulling adjustments, the hand haptically collects additional information to expand the spatial reference created by the resistance of the mass, dragging on the ground. Such haptic information is incorporated into the postural control system

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in order to then reduce body sway [2]. The anchor system has been used in diverse populations such as children [3], healthy adults, adults with intellectual disabilities [4], and older adults [5,6]. The results overall show the body's effective use of haptic information that comes from this system, which improves postural control.

Unstable locomotion in older adults is associated with increased movement amplitude in the frontal plane [7], which has been used as an indication of diminished control in the medio-lateral (ML) direction [7,8]. To compensate for this instability, older adults exhibit various adaptations in their gait patterns, which include reducing step length and speed, as well as increasing step width and double support duration [7,9]. One of the main problems associated with locomotion instability in older individuals is the increased risk for falls [10]. Some studies have shown that, when deficits of a sensory system impair postural control, the addition of sensory information from another input system (*e.g.*, tactile, auditory, or visual biofeedback) can help to improve body stability by reducing trunk sway [11,12].

The addition of haptic information through the light touch of a fingertip on a hard surface consistently has shown a reduction in body sway in older adults [13]. In addition to the use of rigid contact surfaces, non-rigid surfaces, such as a flexible filament, provide additional haptic stimuli and reduce body sway [14]. Light touch on a hard or soft surface also contributes to improve gait stability to individuals walking on the ground or on the treadmill [15–17]. In dynamic tasks, however, the use of the anchor system has been little studied. When walking on a balance beam with obstructed vision, 7-year-old children reduced their trunk sway in the frontal plane with the use of the anchor system [3]. Since the use of the anchor system reduces body sway, it seems important to investigate its impact on gait in older adults, because the majority of falls occur during their performance of dynamic tasks [18]. Thus, one of the purposes of this study was to evaluate the contribution of additional haptic information, provided by the anchor system, on gait stability in older adults who performed an unstable gait task (*i.e.*, tandem gait). We expected that the additional haptic information provided by the anchor system would reduce trunk acceleration in the frontal plane. When an individual uses the anchor system while walking, his or her body orientation relative to the ground is continuously affected by changes in the traction forces of each of the anchor cables, imposed by the anchor tasks. This results in changes in body sway, particularly in the frontal plane. Thus, the anchor system provides additional haptic information about spatial orientation of the body relative to the ground, which helps to maintain the trunk in the vertical position (for detailed explanation see Ref. [2]).

However, the impact of the use of the anchor system should be distinguishable from improvements derived from other input processes such as the effect of just holding a cable while walking or the learning effect that arises from the practice of the task itself. In order to control both effects, we manipulated the anchor task distinctively for two additional groups. One group performed the tandem gait task using an anchor system composed only of the cable and without any mass attached at the end that made contact with the ground. This adapted anchor should not provide any haptic information, since the absence of mass should prevent the cable from tensing. Hence, no additional haptic information on body position relative to the surface support would be available. The other group performed all of the experiments without any type of anchor (*i.e.*, control group).

This distinction is essential to the assessment of short-term effects due to practice with the anchor system. The question is to determine whether or not older adults are able to retain and transfer the performance improvements observed during the practice of postural tasks [19]. In a study with healthy adults and individuals with intellectual disabilities, Mauerberg-deCastro et al. [4] found that only one training session (12 trials) with the anchor system was enough to cause a short-term reduction in body sway during the maintenance of the upright position on a balance beam. Freitas et al. [6] showed that older adults can transfer performance gains obtained during intermittent practice with the anchor system to contexts without the presence of the anchors. However, it has not yet been established whether or not a short period of practice with the anchor system during tandem walking would result in a short-term aftereffect in gait stability, as has been observed in the quiet standing task. Thus, the second purpose of this study was to evaluate the effect of practice with the anchor system on stability of the trunk during a tandem walking task. We expected to find a short-term effect after removal of the anchor system, with participants in the anchor group continuing to exhibit superior performance as compared to their peers in the non-anchor group as well as their peers in the adapted anchor group.

2. Methods

2.1. Participants

Forty-four older adults participated in this study. We randomly assigned them to three groups: control group ($n = 15$), 0 g anchor group ($n = 14$), and 125 g anchor group ($n = 15$). All participants were 65–80 years old, and we matched them by sex, age, height, and body mass. Participants signed a consent form. The local ethics committee approved all procedures used in the study.

Participants of this study were able to understand the verbal instructions necessary to accomplish the tasks without the use of auxiliary devices. Exclusion criteria were: cognitive impairment, stroke, cardiovascular problems, orthopedic injuries, limited mobility of the trunk, severe vision loss or glaucoma, and deficiencies of the vestibular or somatosensory systems. We evaluated these criteria through a questionnaire applied before we began to collect the data. In addition, we assessed the level of physical activity of the participants by applying the modified Baecke Questionnaire for older adults [20], and we used the Mini Mental State Examination to assess the cognitive status of our participants [21]. For functional balance, we applied the MiniBESTest [22]. We assessed tactile sensitivity with the use of an esthesiometer (Sorri®, Bauru, Brazil), composed of 6 monofilaments of nylon, each of increasing diameters [23].

2.2. Procedures

Participants performed all experimental trials on a 5-m GaitRite® system (CIR Systems Inc., Sparta, NJ, USA). This system recognizes the contact of the feet on its surface through the pressure exerted on the sensors distributed inside it and automatically calculates several spatiotemporal gait parameters. In addition, a triaxial accelerometer (Trigno Wireless, Delsys, Boston) was placed on the region of the 7th cervical vertebra to record the acceleration of the torso. We collected the accelerometer data at a frequency of 1500 Hz.

Participants performed the tandem walking task. Tape (2 cm wide) with a contrasting color was placed on the carpet to indicate to the participants the walking direction. We instructed them to step on the tape in all experimental trials. We divided the trials into five blocks of three trials each (fifteen trials per participant). The first three trials were the pre-test phase, in which participants in all three groups performed the tandem walking task without the anchor system (*i.e.*, baseline assessment). Then, participants performed three blocks of trials (totaling nine trials), called the practice phase. In this phase, participants in each group received specific instructions. Participants in the control group performed

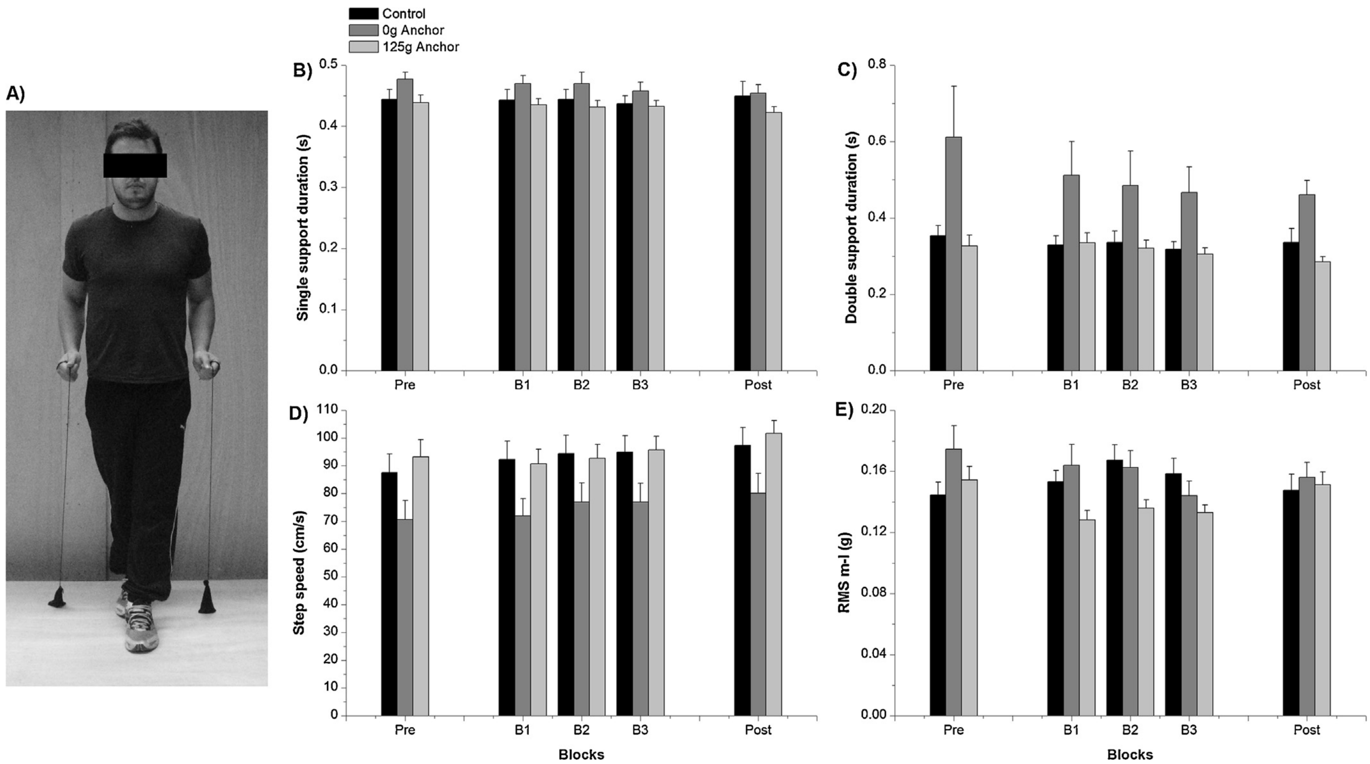

Fig. 1. (A) Picture illustrating the tandem walking with the anchor system. Mean and standard error for the pre-test, practice (Blocks 1 [B1], 2 [B2] and 3 [B3]) and post-test for the single support duration (B), double support duration (C), step speed (D) and RMS m-l (E) to the participants of the three groups.

all of these trials without the anchor system; participants in the 125 g anchor group used the "typical" anchor system; participants in the 0 g anchor group used an adapted anchor. After the practice phase, participants in all three groups performed the post-test phase, which consisted of them performing three trials of the tandem walking task without the anchor system. Participants rested for 30 s between trials and 1 min between blocks.

The "typical" anchor system consisted of a cable with a 125 g mass attached at one end (bird shot secured in a small cloth bag). We asked our participants to hold the cable, one in each hand, while the other end with the 125 g mass rested on the ground. Because of the dynamic nature of the task, participants dragged the anchors as they walked in order to keep them in contact with the ground and to keep the cables taut (Fig. 1). Thus, as the individuals pulled the anchor cables, they sensed changes in tension, which provided haptic information about their body's position relative to the supporting surface. The 0 g anchor condition, in turn, consisted only of the anchor cable, without the attached 125 g mass that, in the other condition, was in contact with the ground. While similar to that of the typical anchor, this condition did not provide haptic information about the body's position relative to the support surface due to an absence of tension on the cable.

2.3. Data analyses

The spatiotemporal parameters of the tandem walking task used in this study were: single support duration, double support duration, and step speed. All of these variables were obtained directly from the GaitRite software. This software provides the values of these variables step-by-step. We used the three intermediate steps of each trial in order to eliminate the initial gait acceleration and the final gait deceleration. For each trial, we calculated the mean value of these three steps. We added these spatiotemporal variables for descriptive purposes only.

For the acceleration analysis, we used the mathematical algorithm proposed by Moe-Nilssen [24] to transform the data into a horizontal-vertical coordinate system. After this transformation, we computed the acceleration root mean square (RMS) in the medial-lateral direction (m-l) in order to identify the variability of the trunk acceleration. To calculate the RMS m-l, three intermediate steps were initially identified on the basis of the vertical acceleration curve. We considered the peak acceleration as the time corresponding to the foot contact with the ground. After identifying the data window that corresponded to the three intermediate steps, we computed the RMS for each window at 1 s, and, afterwards, we calculated the mean value of these windows.

2.4. Statistical analyses

For each gait-dependent variable, we ran three analyses of variance (ANOVA). For the pre- and post-tests, we used 1-way ANOVAs (3 groups). For the practice phase, we ran 2-way ANOVAs (3 groups × 3 blocks) with repeated measures in the second factor. We used a Bonferroni adjusted *post-hoc* analysis. The significance level was .05.

3. Results

We found no difference among groups for the parameters of age, height, and body mass, as well as for the scores in the cognitive evaluation and functional balance test (Table 1). For the level of physical activity, ANOVA 1-way exhibited a main effect for group ($F_{2,43} = 4.001$, $p = .026$). *Post-hoc* analysis identified that the control group was more physically active than the 0 g anchor group ($p = .025$). For tactile sensitivity, we computed the percentage of positive detection of each monofilament number. For both hands, there was a high percentage of positive detection of the lightest monofilament (#1), although the percentage diminished for the

Table 1
Mean and standard deviation (±) of the physical, behavioral, cognitive, and perceptual variables for the participants of each group.

	Control group	0 g Anchor group	125 g Anchor group
n	15	14	15
Age (years)	70.3 ± 4.5	70.4 ± 4.2	70.1 ± 3.7
Height (m)	1.60 ± 0.08	1.60 ± 0.09	1.63 ± 0.08
Body mass (kg)	71.8 ± 12.5	68.8 ± 13.3	71.1 ± 11.5
Physical activity level[a]	6.49 ± 4.20	3.42 ± 2.02	4.51 ± 2.08
Mini-mental state examination[b]	26.3 ± 2.7	26.6 ± 2.7	26.9 ± 2.7
MiniBESTest[c]	27.0 ± 0.8	25.9 ± 1.9	27.1 ± 1.3

Tactile sensitivity	Monofilament number[d]	Control group	0 g Anchor group	125 g Anchor group
Right hand (%)	1	94.3 ± 10.5	99.0 ± 3.8	78.1 ± 30.5
	2	5.7 ± 10.5	1.0 ± 3.8	18.1 ± 28.3
	3	–	–	3.8 ± 8.5
Left hand (%)	1	99.0 ± 3.7	99.0 ± 3.8	82.9 ± 24.9
	2	1.0 ± 3.7	–	13.3 ± 19.8
	3	–	1.0 ± 3.8	3.8 ± 10.1
Right foot (%)	1	37.0 ± 26.3	33.1 ± 29.3	24.2 ± 13.6
	2	18.8 ± 11.6	18.8 ± 11.0	26.7 ± 15.5
	3	38.8 ± 19.9	37.7 ± 20.4	44.8 ± 18.7
	4	4.2 ± 8.3	7.1 ± 8.9	3.6 ± 4.6
	5	1.2 ± 3.2	3.2 ± 5.8	0.6 ± 2.3
Left foot (%)	1	37.6 ± 30.5	36.4 ± 31.3	21.2 ± 14.0
	2	22.4 ± 20.0	17.5 ± 14.9	27.9 ± 15.2
	3	33.9 ± 23.9	34.4 ± 24.5	40.6 ± 16.1
	4	3.6 ± 5.7	7.1 ± 12.4	7.9 ± 11.3
	5	2.4 ± 5.4	4.5 ± 10.5	2.4 ± 4.2

[a] Measured by the Modified Baecke Questionnaire for older adults. Scores close to zero indicates a low level of physical activity.
[b] Scores close to 30 points (maximum punctuation) indicates absence of cognitive deficit.
[c] Scores close to 28 points (maximum punctuation) indicates low risk for falls.
[d] The higher the number of monofilament, the worse is the sensitivity.

125 g anchor group. This suggests a slight reduction in tactile sensitivity of both hands in the 125 g anchor group. For both feet, the detection of the monofilament presence was more distributed among monofilaments #1–#5. Considering that monofilaments #1 and #2 are considered normal for feet, there is a high percentage of normal tactile sensitivity of both feet in all three groups, although some participants presented a reduced sensitivity (equally distributed among the groups).

3.1. Spatiotemporal gait variables

There was no difference among groups for step speed and single support duration in the pre- and post-practice phases. However, for double support duration, ANOVA identified a main effect for group ($F_{2,43} = 4.07$, $p = .024$) in the pre-practice phase. Double support duration was smaller for the 125 g anchor group than for the 0 g anchor group (Fig. 1). For the practice and post-practice phases, ANCOVA (using the pre-test values as the covariate) showed no effect for the double support duration. During the practice phase, simple support duration did not differ among groups and blocks. ANOVA identified main effect of block for step speed ($F_{2,82} = 8.91$, $p = .001$), but no group or interaction effects. Step speed was smaller in block 1 than in blocks 2 ($p = .003$) and 3 ($p = .007$) (Fig. 1).

3.2. Trunk acceleration

RMS did not differ among groups in the pre- and post-practice. ANOVA indicated a main effect for group ($F_{2,41} = 3.77$, $p = .031$) in the practice phase. RMS m-l was lower for the 125 g anchor group as compared to the control group ($p = .049$) (Fig. 1). However, there was no difference between the 125 g anchor group and the 0 g anchor group.

4. Discussion

The purposes of this study were to evaluate the contribution of haptic information provided by the anchor system to walking stability in older adults who performed a tandem walking task and to assess the short-term effect resulting from practice with the anchor system. Relative to the first purpose, the anchor system reduced trunk acceleration in the frontal plane, suggesting an improvement in gait stability. However, no short-term effect was observed for walking, differing from our previous studies of older adults and individuals with intellectual disability who performed a quiet standing task [4,6].

Before we discuss these two main findings, however, we would like to consider some aspects of our sample. There was no difference among groups for both cognitive and balance assessments. The similarity among groups for the MiniBESTest suggests that balance deficit due to aging may not be the cause of the differences in our results. Conversely, tactile sensitivity was slightly impaired in the 125 g anchor group. The increased standard deviation exhibited by the 125 g anchor group suggests a greater diversity amongst the members of this group. Nevertheless, the 125 g anchor group benefited from the use of the anchor system. The fact that the 0 g anchor groups showed better hand tactile sensitivity reinforces the findings of the present study.

In addition, analysis of the results of the physical activity level showed that the 0 g anchor group was less active than the control group, but did not differ from the 125 g anchor group. This effect may be related to the differences in the pre-test values for the double support duration in the 0 g anchor group. The double support is the only part of the gait in which the center of mass is within the limits of the base of support [25]. Thus, the shorter the double support duration, the better the stability control [26], but this gradually becomes impaired with aging. According to Gabell and Nayak [27], double support duration and step width are the first spatiotemporal gait parameters that change with increased instability. The literature shows that physical activity improves walking performance [28]. As the 0 g anchor group exhibited a reduced level of physical activity, we would expect impairment in their walking, although level of physical activity in this group was similar to that of the 125 g anchor group. The values of the double support duration were similar between the 125 g anchor group and the control group, although only the latter differed from the 0 g anchor group.

4.1. Use of the anchor system reduced trunk acceleration in the frontal plane

The main finding of this study was that the additional haptic information provided by the anchor system assisted in reducing trunk acceleration in the frontal plane. This result supports the findings of Calve and Mauerberg-deCastro [3] with 7-year-old children, who reduced trunk sway when they used the anchor system while walking on a balance beam. This result is consistent with other studies showing that the addition of sensory information such as tactile, auditory, and visual biofeedback also improved postural control and stabilization of the trunk in quiet standing and tandem walking tasks [11–14,17]. Bingenheimer et al. [16] also found similar results, demonstrating that the addition of haptic information obtained through light touch via fixed or mobile devices (cane and soft elastic handrail) had a positive effect on spatiotemporal gait parameters as well as on trunk control. In their study, this supplementation of haptic information increased step length and decreased medial-lateral oscillation of the center of mass during walking, as well as the duration of the stance phase in visually restricted conditions. Other studies have also made explicit the positive role of additional haptic information to older adults, but in stationary conditions [5,6,13]. Therefore, the benefits seen with the use of the anchor system in the maintenance of upright posture appear to extend to dynamic tasks like walking in older adults.

The anchor system expands the possibilities for the individual to explore the adjacent environment through the characteristic of perceptual telemodality [29,30]. The concept of telemodality implies the ability of an individual to detect the properties of a distal surface mediated by a cable connection [29]. The anchor system provides the necessary medium for transmitting information from the adjacent support surface to the posture control system. Therefore, the haptic information provided by the anchor system regarding the body's position relative to the ground was used to reduce trunk movement, and, consequently, to improve body stability during walking with a narrow base of support.

4.2. There was no short-term effect after removal of the anchor system

Despite the reduction of trunk acceleration with the use of the anchor system, this effect did not persist in the post-test without the anchor system. It may be that the amount of practice with the anchor system was insufficient to generate a short-term effect. This result contrasts with that of Mauerberg-deCastro et al. [4], who observed a short-term effect after participants practiced with the anchor system in 12 trials while performing a quiet standing task on a balance beam. It is possible that for a dynamic task such as the tandem gait, the amount of practice with the anchor system must be greater than for a less dynamic task. Or, it is possible that the tandem gait is not sufficiently challenging. Perhaps in more challenging tasks the need for the additional information provided by the anchor system is greater, and, as a result, the effects of its use are more evident. Future studies might address practice duration's with more challenging tasks such as walking on a balance beam.

5. Conclusion

The results of this study allow us to infer that the use of the anchor system contributed to a reduction in trunk acceleration in the frontal plane. However, this effect did not persist after removal of the anchors. This suggests that the amount of practice with this tool was insufficient to generate any lasting effect, that the task was not sufficiently challenging, or both.

6. Practical implications

The reduction in trunk acceleration with the use of the anchor system has important implications for real-life situations. It opens the possibility of incorporating the anchor system into intervention programs that focus on balance rehabilitation. From a practical point of view, the anchor system can be easily used in several different contexts, such as physical education classes, clinical practice, and even in home-based intervention programs. Unlike the light touch paradigm that needs a rigid support, stable surface and a long handrail for walking, which hinders its use in more dynamic situations, the anchor system is low cost and allows flexibility of use. Finally, individuals can use the anchor system while walking under varying task demands such as changes in direction and over uneven terrains.

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References

[1] E. Mauerberg-deCastro, Developing an anchor system to enhance postural control, Motor Control 8 (2004) 339–358.
[2] E. Mauerberg-deCastro, R. Moraes, C.P. Tavares, G.A. Figueiredo, S. Pacheco, T. Costa, Haptic anchoring and human postural control, Psychol. Neurosci. 7 (2014) 301–318.
[3] T. Calve, E. Mauerberg-deCastro, Contribuição da percepção háptica no controle postural de crianças [Contribution of the haptic perception in the postural control of children], Motriz: J. Phys. Ed. 11 (2005) 199–204.
[4] E. Mauerberg-deCastro, R. Moraes, D.F. Campbell, Short-term effects of the use of non-rigid tools for postural control by adults with intellectual disabilities, Motor Control 16 (2012) 131–143.
[5] R. Moraes, E. Mauerberg-deCastro, O uso de ferramenta não-rígida reduz a oscilação corporal em indivíduos idosos [The use of a non-rigid tool reduces body sway in older individuals], Motriz: J. Phys. Ed. 15 (2009) 263–272.
[6] M.D.B.Z. Freitas, E. Mauerberg-deCastro, R. Moraes, Intermittent use of an anchor system improves postural control in healthy older adults, Gait Posture 38 (2013) 433–437.
[7] M.A. Schrager, V.E. Kelly, R. Price, L. Ferrucci, A. Shumway-Cook, The effects of age on medio-lateral stability during normal and narrow base walking, Gait Posture 28 (2008) 466–471.
[8] M.J. Hilliard, K.M. Martinez, I. Janssen, B. Edwards, M. Mille, Y. Zhang, M.W. Rogers, Lateral balance factors predict future falls in community-living older adults, Arch. Phys. Med. Rehabil. 89 (2008) 1708–1713.
[9] A.J. Nelson, L.J. Certo, L.S. Lembo, D.A. Lopez, E.F. Manfredonia, S.K. Vanichpong, D. Zwick, The functional ambulation performance of elderly fallers and non-fallers walking at their preferred velocity, NeuroRehabilitation 13 (1999) 141–146.
[10] L.Z. Rubenstein, K.R. Josephson, A.S. Robbins, Falls in the nursing home, Ann. Intern. Med. 121 (1994) 442–451.
[11] M. Dozza, C. Wall, R.J. Peterka, L. Chiari, F.B. Horak, Effects of practicing tandem gait with and without vibrotactile biofeedback in subjects with unilateral vestibular loss, J. Vestib. Res. 17 (2007) 195–204.
[12] M. Dozza, L. Chiari, R.J. Peterka, C. Wall, F.B. Horak, What is the most effective type of audio-biofeedback for postural motor learning? Gait Posture 34 (2011) 313–319.
[13] A.M.S. Baldan, S.R. Alouche, I.M.G. Araujo, S.M.S.F. Freitas, Effect of light touch on postural sway in individuals with balance problem: a systematic review, Gait Posture 40 (2014) 1–10.
[14] J.R. Lackner, E. Rabin, P. DiZio, Stabilization of posture by precision touch of the index finger with rigid and flexible filaments, Exp. Brain Res. 139 (2001) 454–464.

[15] R. Dickstein, Y. Laufer, Light touch and center of mass stability during treadmill locomotion, Gait Posture 20 (2004) 41–47.
[16] K. Bingenheimer, J. Temprado, M. Harnagea, N. Bricot, P. Villani, E. Berton, Effects of a light touch on fixed or mobile supports on gait parameters in visually restricted young adults, Neurosci. Lett. 589 (2015) 176–180.
[17] E. Kodesh, F. Falash, E. Sprecher, R. Dickstein, Light touch and medio-lateral postural stability during short distance gait, Neurosci. Lett. 584 (2015) 378–381.
[18] D.L. Sturnieks, R. St George, S.R. Lord, Balance disorders in the elderly, Clin. Neurophysiol. 38 (2008) 467–478.
[19] K. Van Ooteghem, J.S. Frank, F. Allard, F.B. Horak, Aging does not affect generalized postural motor learning in response to variable amplitude oscillations of the support surface, Exp. Brain Res. 204 (2010) 505–514.
[20] L.E. Voorrips, A.C. Ravelli, P.C. Dongelmans, P. Deurenberg, W.A. Van Staveren, A physical activity questionnaire for the elderly, Med. Sci. Sports Exerc. 29 (1997) 117–121.
[21] S.M.D. Brucki, R. Nitrini, P. Caramelli, P.H.F. Bertolucci, I.H. Okamoto, Suggestions for utilization of the mini-mental state examination in Brazil, Arq. Neuropsiquiatr. 61 (2003) 777–781.
[22] A.C. Maia, F. Rodrigues-de-Paula, L.C. Magalhães, R.L.L. Teixeira, Cross-cultural adaptation and analysis of the psychometric properties of the Balance Evaluation Systems Test and MiniBESTest in the elderly and individuals with Parkinson's disease: application of the Rasch model, Braz. J. Phys. Ther. 17 (2013) 195–217.
[23] L.F. Lehman, M.B. Orsini, A.R. Nicholl, The development and adaptation of the Semmes-Weinstein monofilaments in Brazil, J. Hand Ther. 6 (1993) 290–297.
[24] R.A. Moe-Nilssen, New method for evaluating motor control in gait under real-life environmental conditions. Part 1: the instrument, Clin. Biomech. 13 (1998) 320–327.
[25] D.A. Winter, Human balance and posture control during standing and walking, Gait Posture 3 (1995) 193–214.
[26] N. Mortaza, N.A. Abu Osman, N. Mehdikhani, Are the spatio-temporal parameters of gait capable of distinguishing a faller from a non-faller elderly? Eur. J. Phys. Rehabil. Med. 50 (2014) 677–691.
[27] A. Gabell, U.S. Nayak, The effect of age on variability in gait, J. Gerontol. 39 (1984) 662–666.
[28] A. Tiedemann, C. Sherrington, S.R. Lord, The role of exercise for fall prevention in older age, Motriz: J. Phys. Ed. 19 (2013) 541–547.
[29] P.A. Cabe, Haptic distal spatial perception mediated by strings: haptic looming, J. Exp. Psychol. Hum. Percept. Perform. 37 (2011) 1492–1511.
[30] P.A. Cabe, Haptic distal spatial perception mediated by strings: size at a distance and egocentric localization based on ellipse geometry, Atten. Percept. Psychophys. 75 (2013) 358–374.