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Sensor-based balance training with motion feedback in people with mild cognitive impairment

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Abstract—Some individuals with mild cognitive impairment (MCI) experience not only cognitive deficits but also a decline in motor function, including postural balance. This pilot study sought to estimate the feasibility, user experience, and effects of a novel sensor-based balance training program. Patients with amnestic MCI (mean age 78.2 yr) were randomized to an intervention group (IG, $n = 12$) or control group (CG, $n = 10$). The IG underwent balance training (4 wk, twice a week) that included weight shifting and virtual obstacle crossing. Real-time visual/audio lower-limb motion feedback was provided from wearable sensors. The CG received no training. User experience was measured by a questionnaire. Postintervention effects on balance (center of mass sway during standing with eyes open [EO] and eyes closed), gait (speed, variability), cognition, and fear of falling were measured. Eleven participants (92%) completed the training and expressed fun, safety, and helpfulness of sensor feedback. Sway (EO, $p = 0.04$) and fear of falling ($p = 0.02$) were reduced in the IG compared to the CG. Changes in other measures were nonsignificant. Results suggest that the sensor-based training paradigm is well accepted in the target population and beneficial for improving postural control. Future studies should evaluate the added value of the sensor-based training compared to traditional training.

Clinical Trial Registration: ClinicalTrials.gov; “Virtual reality based balance training in people with mild cognitive impairment”: NCT02214342; https://clinicaltrials.gov/ct2/show/NCT02214342?term=NCT02214342&rank=1

Key words: balance, biofeedback, cognitive impairment, dementia, exercise, exergame, fall risk, fall prevention, interactive, older adults, postural control, wearable sensor.

INTRODUCTION

Mild cognitive impairment (MCI) is a well-recognized risk factor for both Alzheimer disease [1] and functional dependence [2–3], both of which are associated with a

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Abbreviations: ADL = activities of daily living, ANCOVA = analysis of covariance, AP = anterior-posterior, BMI = body mass index, CES-D = Center for Epidemiologic Studies Depression Scale, CG = control group, CoM = center of mass, EC = eyes closed, EO = eyes open, FES-I = Falls Efficacy Scale International, IG = intervention group, MCI = mild cognitive impairment, ML = medial-lateral, MOCA = Montreal Cognitive Assessment, RCT = randomized controlled trial.

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decline in health-related quality of life. Emerging evidence indicates that some individuals with MCI also experience motor dysfunction, including deficits in gait [4–10] and postural control [11–13]. Moreover, some research suggests that motor dysfunction is not a general symptom of MCI but rather is linked to specific MCI subtypes [7,9]. For example, Montero-Odasso et al. demonstrated that participants with the amnestic MCI subtype had poor gait performance, particularly during dual tasks, implying that a motor signature may exist in this MCI subtype [9]. Specific markers of motor dysfunction, such as increased stride-time variability, may also be detected before the prodromal state of MCI, providing further support that a motor phenotype of cognitive decline may exist [14].

While many studies have focused on gait dysfunction and MCI, the association between postural control and MCI has not been as well explored [11–13,15]. Pettersson et al. did not find any differences in traditional balance tests (i.e., Berg Balance Scale) between MCI participants and nondisabled control subjects [15]. However, the Berg Balance Scale reached ceiling effects in both samples (nondisabled and MCI participants), indicating that this scale may not be appropriate for detecting balance differences in more high-functioning populations. In contrast, studies that have used more sophisticated instrumented stabilometry assessments did find increased postural sway during standing balance tasks in amnestic MCI [12] and mixed MCI samples [11,13] compared to their cognitively intact peers. Moreover, postural sway in the medial-lateral (ML) direction was increased in MCI subjects [13], and this sway component was a strong predictor of future fall risk [16–17].

The current evidence on the association between postural balance and MCI is limited to cross-sectional studies that do not prove a causal relation. Nevertheless, the subtle balance changes found in MCI participants compared to their nondisabled peers may indicate a process toward more severe balance disability. Postural control is a key motor function and a hallmark of mobility-related quality of life and independence [18]. The safe performance of balance- and mobility-related activities during daily life requires adequate postural control mechanisms [19].

The subtle balance deficits that seem to exist in participants with MCI encourage the design of tailored balance exercise programs for early intervention. Conventional balance programs, however, may suffer from poor adherence, particularly in an unsupervised setting [20]. Lack of motivation and health complaints have been identified as barriers for exercise training in people with MCI [21]. Further, cognitively impaired persons may have difficulties in correctly executing balance exercises, which include relatively complex movements compared to other exercises such as strength and endurance [22].

With advancements in technology, interactive virtual reality and exergame techniques have been evaluated for balance training [23–24]; these techniques have included positive features such as incentive feedback, enhanced information about motor error to foster motor skill acquisition, and virtually supervised exercise without a personal trainer. To our knowledge, no study has evaluated an interactive technology-based balance training program in people with MCI.

The research project presented in this article focuses on the evaluation of a new interactive training program specifically developed to improve balance [25–26] and builds upon prior work with a community cohort of frail older adults without cognitive impairment [26]. The balance training technology integrates data from motion sensors into a human-computer interface to provide incentive real-time feedback about motion performance during exercising. Such feedback is important to better perceive motor errors during exercise, which are major sources of information for motor learning [27].

The primary aim of this study was to evaluate the feasibility and experience in using the new sensor-based training in a sample of memory clinic patients with clinically confirmed amnestic MCI. To demonstrate proof of concept, we measured the effects on balance after 4 wk of training (twice a week) in comparison to a control group (CG). Based on positive results from previous balance training studies using a comparable training period [26,28–29], we expected to identify improvements in balance. Additionally, we measured effects on gait, cognition, and fear of falling.

METHODS

Study Design

The study was designed as an open-label pilot randomized controlled trial (RCT). The trial was registered at ClinicalTrials.gov (NCT02214342) and was approved by the University of Arizona Institutional Review Board Committee.

Study Population

Individuals were recruited at the Cleo Roberts Memory and Movement Disorders Center of the Banner Sun Health Research Institute (Sun City, Arizona). Recruitment started in July 2014, and follow-up was completed in September 2014. Community-dwelling outpatients with confirmed diagnosis of amnestic MCI according to international established criteria [30] were eligible for the study. Patients meeting MCI criteria were further screened regarding the following exclusion criteria: (1) severe cognitive impairment (Montreal Cognitive Assessment [MOCA] score <20 [31]); (2) nonambulatory or major mobility disorder; (3) other neurological conditions associated with cognitive impairment such as stroke, Parkinson disease, and head injury; (4) any clinically significant psychiatric condition, current drug or alcohol abuse, or laboratory abnormality that would interfere with the ability to participate in the study; (5) severe visual impairment; and (6) unwillingness to participate. Written informed consent was obtained by a board-certified neurologist (MS). A capacity to consent was administered to ensure the subjects were able to understand the consent form and procedures of the study, and all participants were required to personally sign their informed consent. Participants were randomly assigned to intervention group (IG) or CG using computer-generated random numbers. Staff unrelated to the study performed randomization after baseline assessment. **Figure 1** illustrates the progress through the phases of enrollment, allocation, follow-up, and data analysis.

Intervention

Exercise Training Technology

The technology used in this study was specifically developed for measurement and improvement of balance control [25–26,32–34]. It consisted of a 24-inch computer screen, an interactive virtual user interface, and five inertial sensors (LegSys, BioSensics LLC; Cambridge, Massachusetts) including a triaxial accelerometer, gyroscope, and magnetometer for estimation of joint angles and position [33]. Sensor data were acquired and transmitted at a 100 Hz frequency for real-time feedback in a virtual environment. The sensors were mounted in two places (the upper and lower leg) on both legs and on the lower back using elastic straps (**Figure 2**).

Subject Training Procedure

Training was conducted in a separate room in the Cleo Roberts Memory and Movement Disorders Center. The participant stood in front of the screen, which was positioned at eye level. A chair with a backrest (not pictured) was in front of the participant to provide support if required. A supervisor gave instructions about the exercise tasks during the first training session. In subsequent sessions, subjects conducted exercises based on sensor feedback only; however, the supervisor remained with the participant during all sessions to guarantee safety.

Training Protocol

IG participants attended two training sessions per week for 4 wk. Sessions lasted approximately 45 min and included (1) ankle point-to-point reaching tasks and (2) virtual obstacle-crossing tasks. More details on the tasks follow in the next sections. Frequency and duration were determined based on our previous study in cognitively intact older adults, which found positive training effects using an identical training protocol [26]. Given its simple and comprehensive design, the graphical user interface (**Figure 2**) was identical to our previous study in cognitively intact older adults [26]. The interface was designed to be intuitive and easy to navigate and to avoid complex animations that could distract cognitively impaired persons from observing relevant information related to motion performance and motor error.

Ankle Point-to-Point Reaching Task

As shown in **Figure 2(a)–(c)**, the exercise required anterior, posterior, and lateral leaning and partial weight transfer in order to improve postural balance during standing [35]. The ankle reaching task used data from the lower-leg sensors in order to provide visual and auditory feedback during balance exercises. The kinematic of the ankle joint rotation was translated to a linear cursor movement on a computer screen. By rotating the ankle joint, participants had to navigate the cursor from a start circle to a target circle (**Figure 2(a)**). The task was repeated in the opposite direction to complete one cycle. The participant was challenged to navigate rapidly (<1 s) and accurately (in the middle of the circle) from one circle to another. Correct task execution was awarded by visual (circle faded away) and audio (positive sound) feedback. If participants moved too slowly (>1 s), they received visual feedback (circle changed to blue color) to encourage faster movement. To move the cursor forward

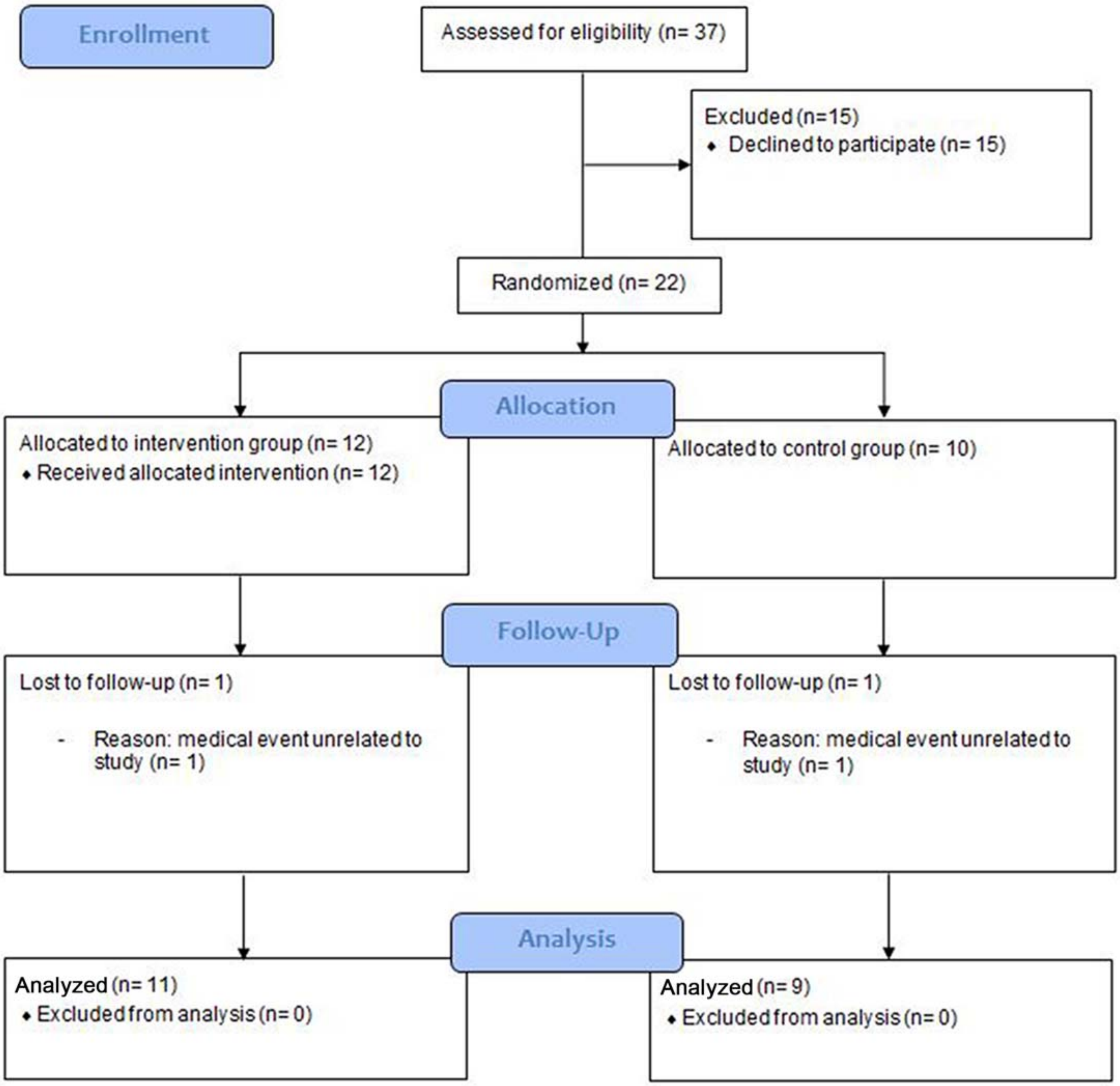

Figure 1.
Consolidated Standards of Reporting Trials flow diagram of progress through the phases of enrollment, allocation, posttesting, and data analysis. (Moher D, Hopewell S, Schulz KF, Montori V, Gøtzsche PC, Devereaux PJ, Elbourne D, Egger M, Altman DG. CONSORT 2010 explanation and elaboration: Updated guidelines for reporting parallel group randomised trials. BMJ. 2010;340:c869).

and backward, participants had to move their hips in an anterior-posterior (AP) direction to generate ankle dorsiflexion or plantar flexion (**Figure 2(a)**). ML hip movement navigated the cursor sideways (**Figure 2(b)**).

One session included 6 blocks, each with 20 cycles of ankle reaching tasks. Blocks 1 and 2 were performed in the AP direction (**Figure 2(a)**). Blocks 3 and 4 combined AP and ML directions (**Figure 2(b)**). To increase

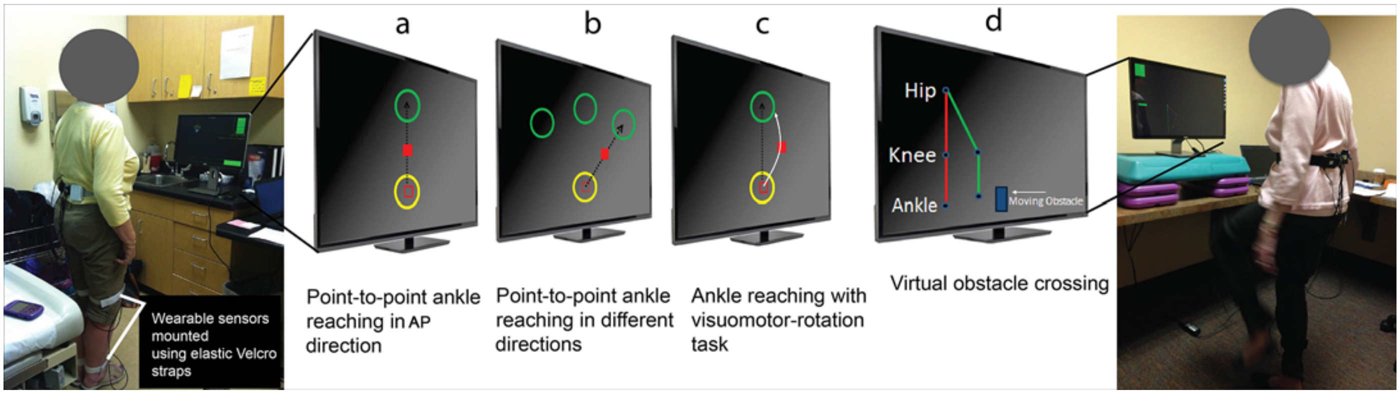

Figure 2.
Illustration of the sensor-based balance training program. **(a)** Ankle reaching task involves navigating a red cursor from a start circle (yellow) to a target circle (green) in a straight line by weight shifting. **(b)** Ankle reaching task is performed in anterior-posterior (AP) and medial-lateral directions. **(c)** Cursor trajectory is rotated by a 20° angle. The user needs to observe the change in trajectory and adjust ankle/hip coordination in order to move the cursor toward the target circle. **(d)** The user has to cross virtual obstacles appearing on the screen. Feedback about lower-limb movement is provided by wearable sensors. This figure has been modified from a previously published figure (Schwenk et al. [26]), which illustrated a similar exergame platform that was used in our previous study to improve balance in community older adults without cognitive impairment.

the task challenge, blocks 5 and 6 were conducted with a visuomotor rotation task [36], and the cursor trajectory on the screen was rotated by 20° (**Figure 2(c)**). The subject had to observe this change in trajectory during the exercise and adjust ankle coordination in order to navigate the cursor in a point-to-point straight line toward the target circle. The visuomotor rotation task focused on training postural adaptation strategies in order to improve postural calibration as described previously [37]. Participants could rest between successive blocks to avoid fatigue.

Virtual Obstacle-Crossing Task

Participants crossed virtual obstacles (boulders) moving on the computer screen from the left to the right side (**Figure 2(d)**). Real-time feedback was provided using a stick figure avatar representing the participant's lower limbs. The avatar replicated the movements of the lower limb, including lifting of the designated foot to the appropriate height to cross an obstacle. Each session included 3 series of virtual obstacle crossing, with 10 repetitions each. The height of the obstacles progressed with each new series (10%, 15%, and 20% of leg length). The subjects received visual and audio feedback, which indicated whether they successfully crossed an obstacle or not. Subjects alternated between right and left legs during an obstacle-crossing sequence. If they forgot the sequence of leg lifting, participants were notified of the mistake by the leg on the screen moving downward instead of upward.

Measurements

Sociodemographics and Clinical Characteristics

At baseline, an interviewer collected participant characteristics including age, sex, body mass index (BMI), cognitive status (MOCA), education (years), activities of daily living (ADL) status (Barthel Index [38]), comorbidities (number of diagnoses), medications (number), depressive symptoms (Center for Epidemiologic Studies Depression Scale, CES-D [39]), pain (visual analog scale ranging from 0 to 10), and history of falls (past year).

User Experience

After the training period, participants described their experience using the sensor-based training technology with an adapted questionnaire originally developed for evaluating the Nintendo Wii balance board (Nintendo; Kyoto, Japan) in older adults [40]. It included 10 Likert-scale questions (i.e., 0 = completely disagree to 4 = absolutely agree) including (1) fun to use, (2) problems during usage, (3) loss of balance, (4) form and design, (5) fear of falling, (6) balance support, (7) helpfulness of sensor-feedback, (8) motion speed, (9) level of difficulty, and (10) safety concerns.

Outcome Measures

Measurements were performed at baseline and after 4 wk using assessments that have been validated in the target population.

Balance

Balance was measured using three wearable sensors (BalanSens, BioSensics LLC) attached to both lower legs and the lower back. Participants were instructed to stand for 30 s with feet close together (without touching), with eyes open (EO) and eyes closed (EC). AP sway (in centimeters), ML sway (in centimeters), and total sway area (in square centimeters) of the center of mass (CoM) was quantified by using validated algorithms [34]. Sample size was estimated for the primary study end point (CoM sway area during EO stance) using results of our previous RCT in older adults without cognitive impairment [26]. In this previous study, CoM sway at follow-up averaged 2.62 ± 1.62 cm^2 in the IG and 1.45 ± 1.01 cm^2 in the CG. Based on an effect size of $d = 0.87$, power of 80 percent, and a significance level of 0.05, a sample size of 24 (12 per group) was required to verify an effect using analysis of covariance (ANCOVA) [41].

Gait

Gait performance was measured using wearable sensors attached to both the left and right upper and lower legs (LegSys, BioSensics LLC). Participants walked 10 m at a normal and a fast pace. Gait speed and variability (coefficient of variation of stride velocity) were calculated using validated algorithms [42].

Fear of Falling

Fear of falling was measured by the Short Falls Efficacy Scale International (FES-I) [43], which consisted of 7 items from the 16-item FES-I. The Short FES-I has good to excellent validity and sensitivity to change in cognitively impaired older adults [44].

Cognitive Performance

Cognitive performance was measured by the MOCA and the Trail Making A and B tests [45].

Statistical Analysis

Results of the user experience questionnaires were calculated as median (range) for each Likert-scale question. Unpaired *t*-tests, Mann-Whitney *U*-tests, and chi-square-tests were used for baseline comparison according to the scale of the investigated variable and the distribution of the data. ANCOVA was employed to compare the effect of the training on outcome parameters at follow-up with adjustment for baseline values [41]. Effect sizes (partial eta squared [ηp^2]) were calculated from ANCOVA. Values from 0.01 to 0.06, 0.06 to 0.25, and above 0.25 indicate small, medium, and large effects, respectively [46].

Using linear regression analyses, we delineated predictive factors of training response for the primary study end point (changes in CoM sway area during standing with EO). Independent variables included age, sex, BMI, ADL status (Barthel Index), cognitive status (MOCA), comorbidities (number of diagnosis), depression (CES-D), fear of falling (Short FES-I), history of falls, and motor performance at baseline (CoM sway, gait speed). Results are reported as β (regression coefficients) and R^2 (coefficient of determination). A p-value ≤ 0.05 was considered to be statistically significant. SPSS statistics, version 22.0 (IBM; Armonk, New York) was used for statistical analysis.

RESULTS

Twenty-two subjects were recruited into the study (**Figure 1**), with $n = 12$ allocated to IG and $n = 10$ to CG. One participant from each group (9.1 percent of the total sample) dropped out during the study period due to acute medical events unrelated to the study intervention. All remaining IG participants ($n = 11$, 92%) completed the eight training sessions. Training was safe despite the participant's advanced age and cognitive impairment. No training-related adverse events occurred.

The mean age of the participants was 78.2 ± 8.7 yr. MOCA score averaged 23.3 ± 2.6 points, indicative of an average score from a population with MCI [31]. All participants were independently living in the community without impairment in ADL status (Barthel Index 98.8 ± 2.8).

Fear of falling was low (Short FES-I <9 points) in 16 participants (72.7%), moderate (9–13 points) in 5 participants (22.7%), and high (≥14 points) in 1 (4.5%) participant. Nine participants (40.9%) reported one or more falls in the last year. Participants' level of depressive symptoms was low (CES-D mean 3.63), with no patient above the cutoff for possible depression (13 points). No significant differences between the IG and CG were found on any baseline variables (**Table 1**).

User Experience

Table 2 shows the results of the user experience questionnaire. The majority of participants expressed that it was fun to use the sensor-based exercise training technology. Likewise, most participants rated the usage, form, and design of the technology positively. They felt safe while using the exercise technology, did not experience fear of falling, never lost their balance while exercising, and did not need balance support while performing the exercises. For the majority of participants, the balance exercises were not difficult to perform and were not going too fast.

Effects of Intervention

Results of baseline and follow-up balance assessments are reported in **Table 3**. With EO, CoM sway was significantly reduced in both directions (AP, ML) in the IG compared to CG after the intervention (p = 0.03 to 0.047). Effect sizes were medium to large (ηp^2 = 0.213 to 0.257).

Fear of falling was significantly reduced in the IG compared to the CG (p = 0.02), with a high effect size (ηp^2 = 0.302).

Change in EC balance (p = 0.18 to 0.28) and gait speed (p = 0.22 to 0.35) were nonsignificant (**Table 3**); however, descriptive results revealed a greater improvement in these

Table 1.
Baseline characteristics of study participants. Data presented as mean ± standard deviation unless otherwise noted.

Characteristic	Intervention (n = 12)	Control (n = 10)	p-Value
Age, yr	77.8 ± 6.9	79.0 ± 10.4	0.76*
Women, No. (%)	7 (58.3)	5 (50.0)	0.70†
BMI, kg/m^2	27.3 ± 3.4	24.8 ± 5.0	0.18*
MOCA, score	23.3 ± 3.1	22.4 ± 3.0	0.48*
Education, yr	14.2 ± 2.3	15.9 ± 2.7	0.13*
Barthel Index (ADL), score	99.6 ± 1.4	98.0 ± 3.5	0.21‡
CES-D, score	3.6 ± 3.2	3.0 ± 2.8	0.64*
Short-FES-I, score	8.8 ± 4.3	8.7 ± 1.8	0.99*
Diagnoses, No.	2.5 ± 1.6	3.5 ± 2.1	0.22*
Prescriptions, No.	4.0 ± 2.0	6.3 ± 3.9	0.09*
Visual Analog Pain Scale (0–10), score	1.3 ± 1.9	0.8 ± 1.1	0.43‡
History of Falls in Last Yr, No. (%)	6 (50)	3 (30)	0.34†
Walking Aid User, No. (%)	1 (8.3)	1 (10)	0.89†
Community Dwelling, No. (%)	12 (100)	10 (100)	>0.99†

*p-value for t-test for differences between intervention and control group.
†p-value for chi-square for differences between intervention and control group.
‡p-value for Mann-Whitney U-test for differences between intervention and control group.
ADL = activities of daily living, BMI = body mass index, CES-D = Center for Epidemiological Studies Depression Scale, FES-I = Falls Efficacy Scale International, MOCA = Montreal Cognitive Assessment, No. = number.

Table 2.
Results of the user experience questionnaire.

Question	Median	Range
1: It was fun to use the sensor-based balance exercise technology.	4	2–4
2: Usage of the technology was possible without problems at any time.	4	1–4
3: I never lost my balance while using the exercise technology.	3	0–4
4: The form and design of the technology are optimal for me.	4	2–4
5: I was afraid to tumble or to fall during the exercise.	0	0–3
6: I required balance support while conducting the exercises.	1	0–4
7: Thanks to the sensor-feedback, I could quickly learn all exercises.	3	0–4
8: I feel that the exercises were going too fast for me.	0	0–1
9: Some of the movements were difficult to perform.	0	0–4
10: I felt safe using the exercise technology.	4	0–4

Note: Response categories: 0 = disagree completely, 1 = disagree moderately, 2 = neutral, 3 = agree moderately, 4 = agree absolutely.

Table 3.
Effects of the sensor-based balance training on outcome parameters. Data presented as mean ± standard deviation unless otherwise noted.

Parameters	Control			Intervention			*p*-Value†	Effect Size‡
	Baseline *n* = 9	Follow-Up *n* = 9	% Change*	Baseline *n* = 11	Follow Up *n* = 11	% Change*		
Balance: EO								
CoM sway, cm^2	2.66 ± 4.03	5.56 ± 10.74	−109.0	2.50 ± 2.53	1.09 ± 0.71	56.4	0.04	0.224
ML CoM sway, cm	1.62 ± 0.90	1.87 ± 1.65	−15.4	1.60 ± 0.76	1.05 ± 0.44	34.4	0.047	0.213
AP CoM sway, cm	1.22 ± 1.05	1.69 ± 1.67	−38.5	1.31 ± 0.74	1.00 ± 0.42	23.7	0.03	0.257
Balance: EC								
CoM sway, cm^2	5.64 ± 11.19	5.43 ± 8.20	3.7	2.22 ± 2.78	2.05 ± 3.07	7.7	0.28	0.073
ML CoM sway, cm	2.20 ± 1.82	2.09 ± 1.30	5.0	1.35 ± 0.60	1.18 ± 0.69	12.6	0.19	0.104
AP CoM sway, cm	1.48 ± 1.54	1.80 ± 1.55	−21.6	1.34 ± 0.97	1.30 ± 1.02	2.9	0.18	0.110
Gait: Habitual Walking								
Speed, m/s	1.06 ± 0.17	1.10 ± 0.20	3.8	0.98 ± 0.22	1.05 ± 0.22	7.1	0.35	0.052
Stride-Time Variability, CV	3.48 ± 1.19	2.75 ± 1.35	21.0	3.13 ± 1.11	2.72 ± 1.11	13.1	0.78	0.005
Gait: Fast Walking								
Speed, m/s	1.44 ± 0.22	1.34 ± 0.37	−6.9	1.39 ± 0.35	1.43 ± 0.34	2.9	0.22	0.087
Stride-Time Variability, CV	3.11 ± 1.23	3.95 ± 4.13	−27.0	3.40 ± 1.30	3.55 ± 1.38	−4.4	0.83	0.003
Fear of Falling: Short FES-I, score	8.8 ± 1.9	9.8 ± 1.4	−11.4	9.0 ± 4.2	8.2 ± 1.4	8.9	0.02	0.302
Cognitive Performance								
MOCA, score	23.2 ± 1.5	25.3 ± 1.9	+8.6	23.3 ± 3.1	23.7 ± 3.9	+1.7	0.13	0.122
Trail A, s	42.4 ± 20.0	45.1 ± 21.0	−6.3	51.8 ± 24.3	46.0 ± 14.1	+11.2	0.69	0.009
Trail B, s	98.9 ± 43.0	99.8 ± 39.5	−1.0	149.2 ± 89.5	155.6 ± 101.3	−4.3	0.74	0.006

*Positive values indicate improvement.
†*p*-values from ANCOVA comparing the effect of the intervention on outcome parameters at follow-up adjusting for baseline values.
‡Effect size (partial eta squared) from ANCOVA.
ANCOVA = analysis of covariance, AP = anterior-posterior, CoM = center of mass, CV = coefficient of variation, EC = eyes closed, EO = eyes open, FES-I = Falls Efficacy Scale International, ML = medial-lateral, MOCA = Montreal Cognitive Assessment.

outcomes in terms of the pre- to postintervention change in the IG (range 2.9% to 12.6%) compared to the CG (range –21.6% to 5.0%), yielding medium effect sizes (range $\eta p^2 = 0.052$ to 0.110).

No intervention effects were obtained for stride-time variability (p = 0.78 to 0.83) and cognitive performance (p = 0.13 to 0.74).

Predictors of Training Response

Regression analysis showed that low baseline balance performance (higher CoM sway area, EO) was associated with greater improvement in the primary study end point in the IG (pre- to postintervention reduction in CoM sway area, EO: $\beta = -0.961$, $R^2 = 0.924$, $p < 0.001$) (**Figure 3**). Further, we identified an association between fall history and pre- to postintervention improvement in ML balance control in the IG. Those patients who had fallen in the previous year showed more improvement in ML CoM sway measured in the EO condition ($\beta = -0.661$, $R^2 = 0.428$, p = 0.03). Other baseline parameters did not significantly predict training response (p = 0.07 to 0.99).

DISCUSSION

To our knowledge, this is the first study to evaluate a sensor-based interactive balance training program in people with MCI. Results of this pilot study show that the proposed balance training program was feasible and well accepted in patients with MCI. Positive effects on balance and fear of falling suggest benefits of the training for improving postural control.

The high adherence rate and the positive user feedback suggest that the training program was well accepted

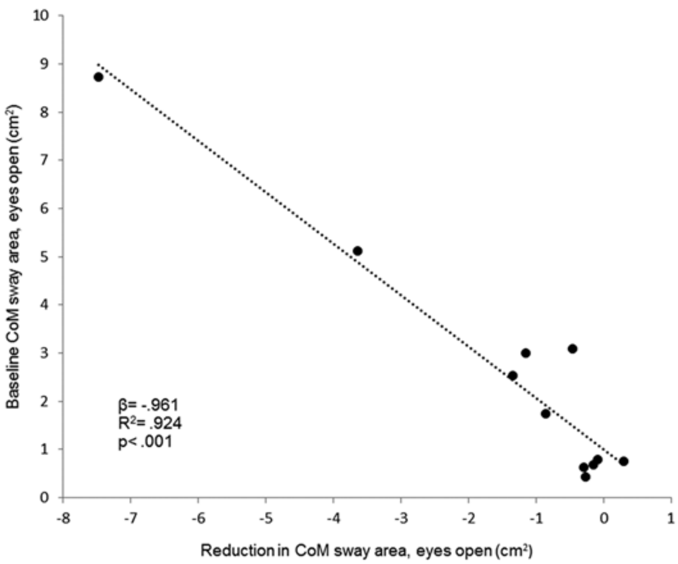

Figure 3.
Association between baseline balance performance and training benefit. Patients with higher center of mass (CoM) sway at baseline benefited more from the balance training as reflected by a greater reduction in CoM sway after the intervention period.

by the study participants, although this needs to be confirmed in a larger trial. Previous training studies have reported inappropriate intensity (either too high or too low) or general complaints related to those programs as major reasons for discontinuing exercising in people with MCI [21]. In contrast, users in this pilot study judged the task demand as appropriate and did not feel overtaxed or concerned about safety during training. A previous study used the same questionnaire for evaluating older adult's experience in using the Wii Fit platform [40]. While this earlier study found comparable results for fun to use (median 4, our study 4), the Wii users were more afraid of falling during training (median 3, our study 0) and found the technology less helpful for quickly learning the exercises (median 1, our study 3). These differences may suggest better perceived support through sensor-feedback in mastering balance exercises as compared to platform-based feedback. Also, the simplistic design of the graphical user interface used in our study may have allowed users to focus on the balance tasks instead of being distracted by other animations as reported for off-the-shelf video games [40].

Commercial exergaming approaches have been demonstrated to be feasible and beneficial for improving balance [47–48]; however, they have limitations in impaired populations at high risk of falling. Force platforms such as the Nintendo Wii restrict the base of support during exercising, which may cause falls during training [40,49]. In contrast, the sensor-based system used in the present study allowed participants to exercise on the ground in a natural stance position. Unlike camera-based exergame systems like the Microsoft Kinect (Microsoft; Redmond, Washington), our wearable sensor system did not require a continuous unobstructed sightline, and we were able to place a chair in front of the participant as a mechanism to prevent falls during training. This safety feature is particularly important during balance training in patients with MCI who have increased fall risk.

Reduced postural sway in the IG after the training may suggest a positive effect of the balance training. However, it should be noted that this pilot concept study did not compare the interactive balance training to another training program. A study with an active CG is needed to more clearly determine whether the effects found in this study were related to our proposed training. Nevertheless, results of this study are promising and function as an initial step toward designing a sensor-based training approach for the target population.

Similar to other studies in people with MCI [13], our participants had higher ML than AP baseline sway, indicating increased risk of falling. In contrast, nondisabled subjects show larger postural sway in the AP than the ML direction, mainly because of structural mechanisms of the ankle and hip joints [50–51]. Interestingly, our results show that ML sway was substantially reduced by 34.4 percent after the training period. These findings imply that MCI-specific deficits in postural balance, such as ML instability [13], may be improved using our training program.

Results from the regression analysis suggest that participants with lower baseline balance performance had greater improvements in balance. These findings are in line with previous studies in persons with cognitive impairment that showed those with the lowest performance benefited most from exercise training [52]. Further, we found that those patients who had fallen in the previous year improved more in ML CoM sway measured in the EO condition. Greater ML balance gains found in those who had fallen seemed to be related to their lower baseline ML balance performance. Results suggest that fallers had lower baseline ML balance performance than nonfallers, indicated by a higher ML COM sway (1.88 ±

0.71 cm vs 1.26 ± 0.71 cm, respectively), although the difference was nonsignificant in the small IG sample (p = 0.19). Because of greater baseline balance deficits, fallers may have had more room to improve balance during the training intervention.

Positive effects on fear of falling suggest that the effect of the balance training may have led to an increased self-efficacy at avoiding falls during ADL. No significant improvements were obtained for balance assessed during the more challenging EC condition or for gait assessment. However, our pilot study's sample size was calculated on a reduction in sway during EO standing and may have been too small to verify significant effects in other outcomes.

Some studies of motoric training have found associated improvement in cognitive performance [53]. In the present study, the lack of effect on cognitive performance is most likely related to the relatively short training period and lack of training specificity for improving cognitive performance [54]. Based on the positive findings of this study, we are further developing the presented training approach by combining balance tasks with cognitive tasks displayed on the computer screen for specific training of dual-task performance [55].

The present pilot study has several limitations. First, the sample size was small. Second, we had relatively few sessions per week with a fairly limited duration of training. Third, the CG was not involved in any other form of exercise. Fourth, the sustainability of training effects found in our study remains unclear. Subsequent research with follow-up at 3 to 6 mo and an active CG (i.e., conventional balance training, commercial exergames) is required to further evaluate the potential of our training. Finally, the positive results on user experience need to be interpreted with caution. Our participants may have had extra motivation to perform the exercise because they were participating in the study. Further, we assisted the participants in taking on and off the sensor straps and setting up the balance training software. Assistance was required because we used a prototype that is not yet designed for fully unsupervised training. Another area for future work will be to determine the extent to which the positive perceptions of the system translate into long-term compliance. Based on the positive results from this study, we are further developing the exercise technology to include a more user-friendly software application and Bluetooth technology for automated connection of sensors with a computer for autonomous in-home training.

CONCLUSIONS

The positive results of this pilot study are a first step toward evaluating a new balance training paradigm specifically designed for improving balance in people with MCI. Current findings may help to inform tailored interventions integrating wearable sensors for interactive balance training in a clinical or home environment.

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