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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 years) were randomized to intervention group (IG, n= 12) or control group (CG, n= 10). The IG underwent balance training (4 weeks, twice a week) including weight shifting and virtual obstacle crossing. Real-time visual/audio lower-extremity motion feedback was provided using wearable sensors. The CG received no training. User experience was measured by a questionnaire. Post intervention effects on balance (center of mass sway during standing with eyes open 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 (eyes open, p=.041) and fear of falling (p=.015) were reduced in IG compared to CG. Changes in other measures were non-significant. 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. Trial registration: www.clinicaltrials.gov .

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Statement of responsibility

Michael Schwenk: Concept and design, study coordination, analysis and interpretation of data, preparation of manuscript. Marwan Sabbagh: Patient recruitment, clinical supervision, study coordination. Ivy Linn: Acquisition of data. Pharah Morgan: Acquisition of data. Gurtej Singh Grewal: Data analysis. Jane Mohler: Study coordination, Concept and design. David W. Coon: Concept and design. Bijan Najafi: Concept and design. All authors contributed in critically revising the manuscript and have given final approval of the version to be published.

Keywords

balance; biofeedback; cognitive impairment; dementia; exercise; fall risk; interactive; older adults; postural control; wearable sensor

Introduction

Mild cognitive impairment (MCI) is a well-recognized risk factor for both Alzheimer's Disease (1) and functional dependence (2, 3), both of which are associated with a 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 under 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). Petterson et al. did not find any difference in traditional balance tests (i.e. Berg-Balance-Scale) between MCI participant and healthy controls (15). However, the Berg-Balance-Scale reached ceiling effects in both samples (healthy and MCI participants), indicating that this scale may not be appropriate for detecting balance differences in more high functioning populations. In contrast, studies which 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 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 which do not prove a causal relation. Nevertheless, the subtle balance changes found in MCI participants compared to their healthy peers may indicate a process towards 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 which 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 rather complex movements as 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 paper 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 elders 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 weeks of training (twice a week) in comparison to a control group. 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 RCT. The trial was registered at www.clinicaltrials.gov () and was approved by the University of Arizona Institutional Review Board Committee.

Study Population

Individuals were recruited at the Memory Disorders Clinic of the Banner Sun Health Research Institute (Sun City, AZ). 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) non-ambulatory or major mobility disorder, 3) other neurological conditions associated with cognitive impairment such as stroke, Parkinson's 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 control group (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 screening, 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 (LegSysTM, BioSensics LLC, MA, USA) including a tri-axial 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 on 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 Banner Sun Health Memory Clinic. The participant stood in front of the screen which was positioned at eye-level. A chair with backrest 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 four weeks. Sessions lasted approximately 45 minutes and included: 1) ankle point-to-point reaching tasks and 2) virtual obstacle crossing tasks (described below). Frequency and duration was 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 2a). The task was repeated in the opposite direction to complete one cycle. The participant was challenged to navigate rapidly (< 1 second) and accurately (in middle of 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 second), they received visual feedback (circle changed to blue color) to encourage faster movement. To move the cursor forward and backward, participants had to move their hips in an anterior-posterior direction to generate ankle dorsi-flexion or plantar-flexion (Figure 2a). Medial-lateral hip movement navigated the cursor sideward (Figure 2b).

One session included 6 blocks, each with 20 cycles of ankle reaching tasks. Blocks 1 and 2 were performed in anterior-posterior direction (Figure 2a). Blocks 3 and 4 combined anterior-posterior and medial-lateral directions (Figure 2b). To increase 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 a 20° angle (Figure 2c). 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 towards the target circle. The visuomotor rotation task focused on training postural adaptation strategies in order to improve postural calibration as described earlier (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 2d). Real-time feedback was provided using a stick figure avatar representing the participant's lower extremities. The avatar replicated the movements of the lower extremity including lifting of the designated foot to the appropriate height to cross an obstacle. Each session included three series of virtual obstacle crossing, with ten 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, gender, BMI, cognitive status (MOCA), education (years), ADL-status (Barthel Index (38)), comorbidity (number of diagnoses), medications (number), depressive symptoms (Center for Epidemiologic Studies Depression Scale, CES-D, (39)), pain (Visual Analogue Scale ranging from 0–10), and history of falls (past year).

User experience: After the training period, participants described their experience in using the sensor-based training technology with an adapted questionnaire originally developed for evaluating the Wii balance board in older adults (40). It included ten 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 four weeks using assessments that have been validated in the target population.

Balance: Balance was measured using three wearable sensors (BalanSens™, BioSensics, MA, USA) attached to both lower legs and the lower back. Participants were instructed to stand for 30 seconds with feet close together (without touching), with eyes open (EO), and eyes closed (EC). Anterior-posterior (AP, cm) sway, medial-lateral (ML, cm) sway, and total sway area (cm^2) of the center of mass (CoM) was quantified as using validated algorithms (34). Sample size was estimated for the primary study endpoint (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%, and a significance level of .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, MA, USA). Participants walked ten meters at normal and 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 (Short-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. Analysis of covariance (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 endpoint (changes in CoM sway area are during standing with EO). Independent variables included age, gender, BMI, ADL-status (Barthel), cognitive status (MOCA), comorbidity (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$\leq$ 0.05 was considered to be statistically significant. SPSS statistics 22.0 (IBM, Armonk, NY, USA) was used for statistical analysis.

Results

Thirty-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% 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%) completing 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 years. 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 mean: 98.8 ± 2.8).

Fear of falling was low (Short-FES-I< 9 points) in 16 participants (72.7%), moderate (9–13 points) in five participants (22.7%), and high (≥ 14 points) in one (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 cut-off 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 the 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= .027–.047). Effect sizes were moderate to large (ηp^2= .213–.257).

Fear of falling was significantly reduced in the IG compared to the CG (p= .015), with high effect size (ηp^2= .302).

Change in EC balance (p= .178–.333) and gait speed (p= .222–.833) were non-significant (Table 3); however descriptive results revealed a greater improvement in these outcomes in terms of the pre to post change in the IG (range: 2.9%–12.6%) compared to the CG (range: −21.6%−5.0%) yielding medium effect sizes (range ηp^2= .052–.110).

No intervention effects were obtained for stride time variability (p= .780–.833) and cognitive performances (p= .132–.738).

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 endpoint in the IG (pre- to post reduction in CoM sway area, EO: β= −.961, R^2= .924, p< .001) (Figure 3). Further, we identified an association between fall history and pre-to-post 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 (−.661, R^2= .428, p= .027). Other baseline parameters did not significantly predict training response (p=.066–.989).

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 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 to fall during training (median 3; our study 0) and expressed less support from the technology for conducting exercises (median 1; our study 3) compared to our study participants. 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 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, 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 MCI patients with 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 control group is needed to more clearly determine if 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 towards designing a sensor-based training approach for the target population.

Similar to other studies in people with MCI (13), our participants had higher ML baseline sway compared to AP, indicating increased risk of falling. In contrast, healthy subjects show larger postural sway in the AP as compared to the ML direction, mainly due to structural mechanism of ankle and hip joints (50, 51). Interestingly, our results show that ML sway was substantially reduced by 34.4% 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 cognitively impaired persons 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 fallers seemed to be related to their lower baseline ML balance performance. Results suggest that fallers had lower baseline ML balance performance indicated by a higher ML COM sway (1.88 ± 0.71 cm) compared to non-fallers (1.26 ± 0.71 cm), although non-significant in the small IG sample ($p= .192$). Because of greater baseline balance deficits, fallers may have had more room for improving 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 activities of daily living. No significant improvements were obtained for balance assessed during the more challenging EC condition as well as 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 for verifying 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 effects 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 and we had relatively few sessions per week and a fairly limited duration of training. Second, the control group was not involved in any other form of exercise. Third, the sustainability of training effects found in our study remains unclear. Subsequent research with follow-up at 3–6 months and an active control group (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 towards 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.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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Abbreviations:

ADL	activities of daily living
ANCOVA	analysis of covariance
AP	anterior-posterior
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
ML	medial-lateral
MCI	mild cognitive impairment

MOCA	Montreal Cognitive Assessment
RCT	randomized controlled trial

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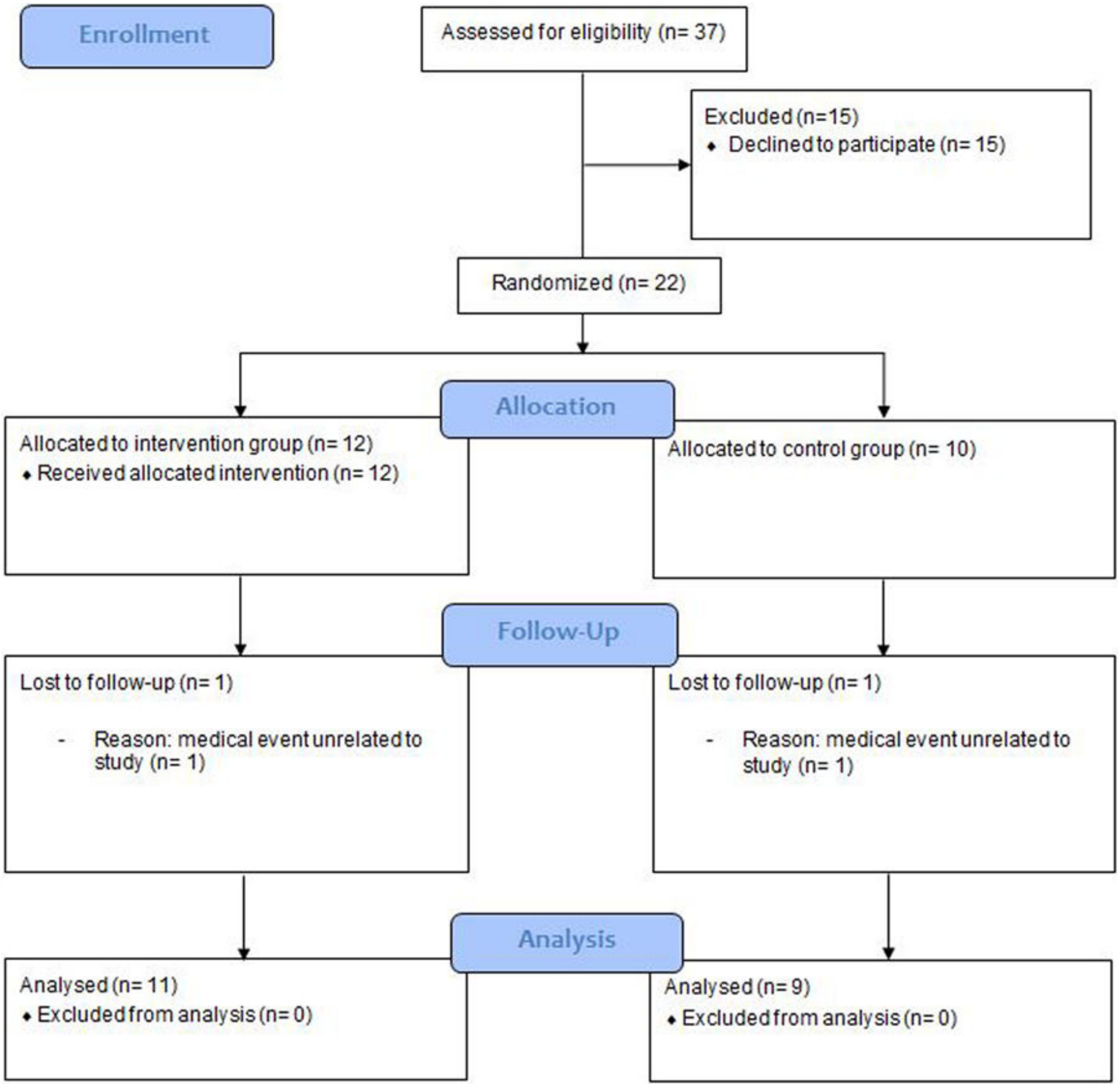

Figure 1. CONSORT flow diagram of progress through the phases of screening, enrolment, allocation, post-testing, and data analysis

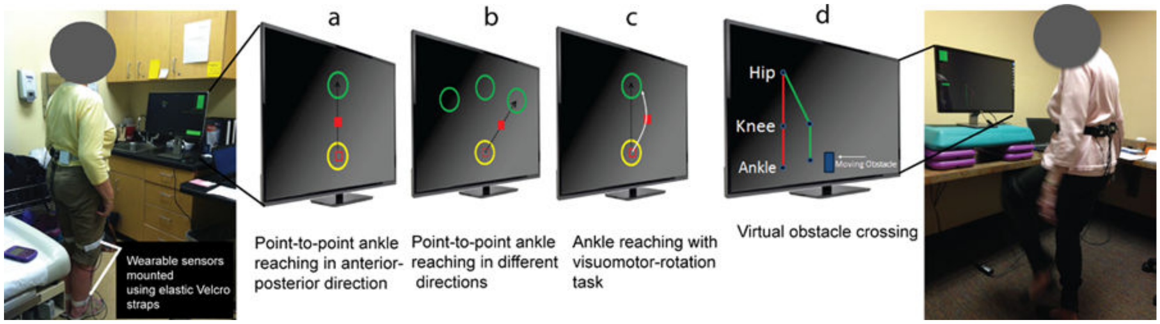

Figure 2. An illustration of the sensor-based balance training program.

a: The 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:** The ankle reaching task is performed in anterior-posterior and medial-lateral direction. **c:** The 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 towards the target circle. **d:** The user has to cross virtual obstacles appearing on the screen. Feedback about lower extremity movement is provided by wearable sensors.

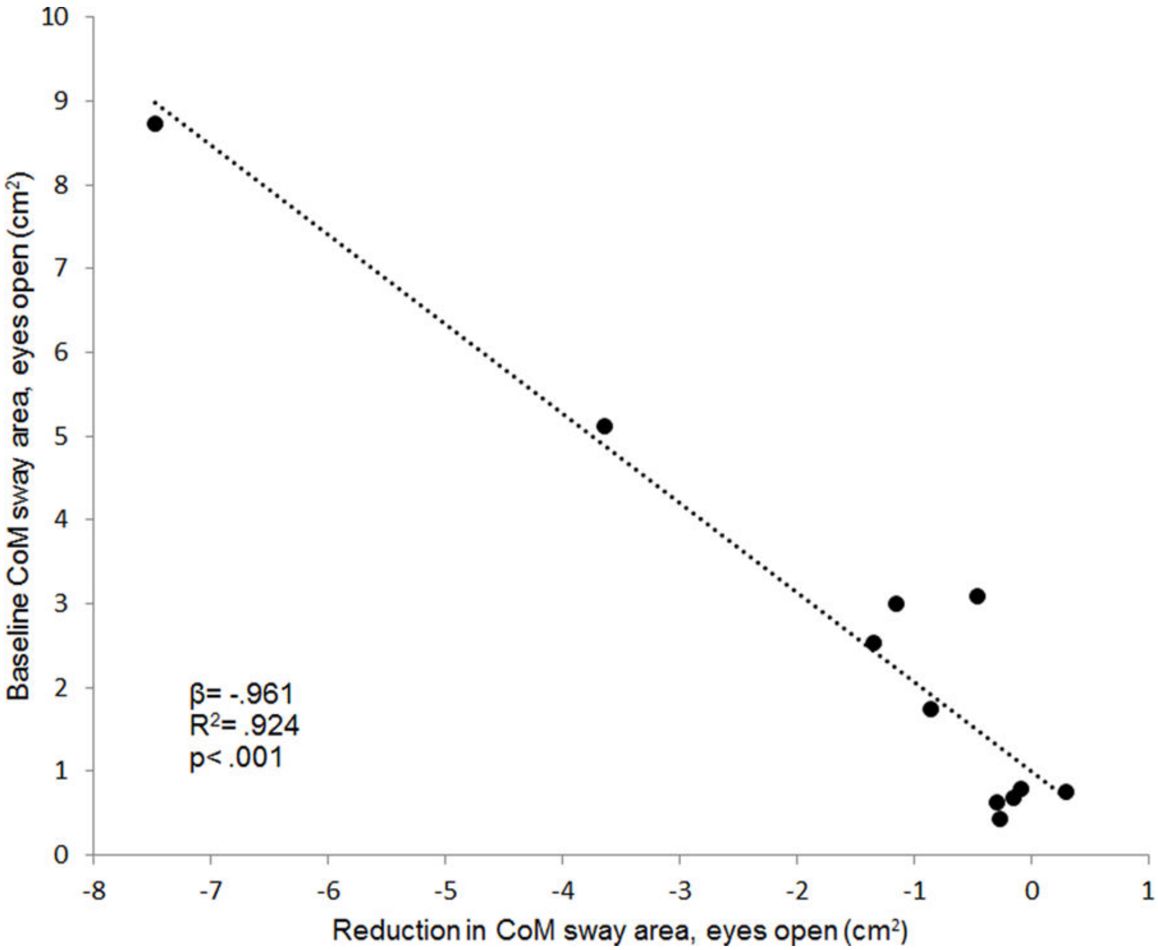

Figure 3. Association between baseline balance performance and training benefit.

Patients with higher CoM sway at baseline benefited more from the balance training as reflected by a greater reduction in CoM sway after the intervention period.

Table 1.

Baseline characteristics of study participants

Characteristic	Intervention (n = 12)	Control (n = 10)	P-value
Age, years	77.8 ± 6.9	79.0 ± 10.4	.756[a]
Women, number	7 (58.3)	5 (50.0)	.696[b]
BMI, kg/m^2	27.3 ± 3.4	24.8 ± 5.0	.182[a]
Montreal Cognitive Assessment, score	23.3 ± 3.1	22.4 ± 3.0	.484[a]
Education, years	14.2 ± 2.3	15.9 ± 2.7	.134[a]
Barthel Activities of Daily Living, score	99.6 ± 1.4	98.0 ± 3.5	.206[c]
CES-D scale, score	3.6 ± 3.2	3.0 ± 2.8	.638[a]
Short Fall Efficacy Scale, score	8.8 ± 4.3	8.7 ± 1.8	.986[a]
Diagnoses, number	2.5 ± 1.6	3.5 ± 2.1	.218[a]
Prescriptions, number	4.0 ± 2.0	6.3 ± 3.9	.090[a]
Visual Analogue Pain Scale (0–10), score	1.3 ± 1.9	0.8 ± 1.1	.432[c]
History of falls in the last year, number of participants	6 (50)	3 (30)	.342[b]
Walking aid user, number	1 (8.3)	1 (10)	.892[b]
Community dwelling, number	12 (100)	10 (100)	1.000[b]

Data are mean ± standard deviation or number (%); P- values for [a]t-test, [b]Chi-square, and [c] Mann-Whitney-U-Test applied to test for differences between intervention and control group; CES-D, Center for Epidemiological Studies Depression Scale.

Table 2.

Results of the user experience questionnaire

Question	Median	Range
Q1: It was fun to use the sensor-based balance exercise technology.	4	2–4
Q2: Usage of the technology was possible without problems at any time.	4	1–4
Q3: I never lost my balance while using the exercise technology.	3	0–4
Q4: The form and design of the technology are optimal for me.	4	2–4
Q5: I was afraid to tumble or to fall during the exercise.	0	0–3
Q6: I required balance support while conducting the exercises.	1	0–4
Q7: Thanks to the sensor-feedback, I could quickly learn all exercises.	3	0–4
Q8: I feel that the exercises were going too fast for me.	0	0–1
Q9: Some of the movements were difficult to perform.	0	0–4
Q10: I felt safe using the exercise technology.	4	0–4

Answer 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

Parameters	Control Group			Intervention Group			P value[b]	Effect size[c]
	Baseline n=9	Follow Up n=9	% change[a]	Baseline n=11	Follow Up n=11	% change[a]		
Balance - Eyes open								
CoM sway, area, cm	2.66 ± 4.03	5.56 ± 10.74	−109.0	2.50 ± 2.53	1.09 ± 0.71	56.4	.041	.224
CoM sway, ML, cm	1.62 ± 0.90	1.87 ± 1.65	−15.4	1.60 ± 0.76	1.05 ± 0.44	34.4	.047	.213
CoM sway, AP, cm	1.22 ± 1.05	1.69 ± 1.67	−38.5	1.31 ± 0.74	1.00 ± 0.42	23.7	.027	.257
Balance - Eyes closed								
CoM sway, area, cm	5.64 ± 11.19	5.43 ± 8.20	3.7	2.22 ± 2.78	2.05 ± 3.07	7.7	.277	.073
CoM sway, ML, cm	2.20 ± 1.82	2.09 ± 1.30	5.0	1.35 ± 0.60	1.18 ± 0.69	12.6	.192	.104
CoM sway, AP, cm	1.48 ± 1.54	1.80 ± 1.55	−21.6	1.34 ± 0.97	1.30 ± 1.02	2.9	.178	.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	.349	.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	.780	.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	.220	.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	.833	.003
Fear of falling								
Short-FESI, score	8.8 ± 1.9	9.8 ± 1.4	−11.4	9.0 ± 4.2	8.2 ± 1.4	8.9	.015	.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	.132	.122
Trail A, sec	42.4 ± 20.0	45.1 ± 21.0	−6.3	51.8 ± 24.3	46.0 ± 14.1	+11.2	.689	.009
Trail B, sec	98.9 ± 43.0	99.8 ± 39.5	−1.0	149.2 ± 89.5	155.6 ± 101.3	−4.3	.738	.006

Data are presented as mean ± standard deviation

[a] positive values indicate improvement

[b] P-values from ANCOVA comparing the effect of the intervention on outcome parameters at follow-up adjusting for baseline values

[c] Effect size eta squared from ANCOVA

CoM, center of mass; ML, mediolateral; AP, anteriorposterior; CV, coefficient of variation; MOCA, Montreal Cognitive Assessment; FES-I, Falls Efficacy Scale International.