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Decline in sensorimotor systems explains reduced falls self-efficacy

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ABSTRACT

Physical performance including balance tasks is one of the main factors explaining the variance in falls self-efficacy in older adults. Balance performance is often measured by use of gross assessment scales, which assess the result of integration of all systems involved in postural control. We aimed to investigate which measurements of postural control correlate to falls self-efficacy scores as measured by the FES-I instrument, and which sensory and motor systems best explain them.

A cross sectional study was designed, in which 45 older adults performed quiet stance and limits of stability trials during which their center of pressure (CoP) excursion was recorded. Falls self-efficacy was measured using the Falls Efficacy Scale – International. Eyesight, vestibular function, proprioception, reaction time and strength were also measured. Hierarchical orthogonal projection of latent structures was used to model FES-I with the CoP trials and then with the sensory and muscle function data.

Fes-I could be explained to 39%, with the eyes open trials and the limits of stability trials loading the heaviest. The base model could be explained to 40% using the sensory and muscle function data, with lower limb strength, leg proprioception, neck proprioception, reaction time and eyesight loading the heaviest.

1. Introduction

Human postural control acts in order to maintain orientation and equilibrium, i.e. balance. To create balance, the central nervous system integrates several modalities of sensory information – from visual, vestibular, and somatosensory receptors – and creates coordinated motor actions and reactions [Horak, 2006; Shumway-Cook, 2007]. This sensorimotor integration takes place in relation to the demands brought about by the motor task in progress and the environment in which it takes place.

Reduced falls self-efficacy is a form of fall-related concern and causes a dramatic decline in physical activity [Choi et al., 2017; Deshpande et al., 2008; WHO, 2002]. These concerns can develop before an actual fall ever happens [Davis et al., 2009; Maki, 1997; Yardley and Smith, 2002] and can be seen as both a cause and a consequence of falls [Lavedan et al., 2018]. It has been hypothesised that fall risk and fall-related concern are two separate consequences of decreased postural control [Hadjistavropoulos et al., 2011]. The latter was confirmed in our previous study showing that physical performance including balance tasks is one of the main factors explaining the variance in falls self-efficacy in older adults, as measured by the Falls Efficacy Scale – International (FES-I) [Pauelsen et al., 2017]. When balance performance is measured, this is often done by use of gross assessment scales, such as the Short Physical Performance Battery (SPPB) [Kapan et al., 2017; Pauelsen et al., 2017]. These types of measurements assess the result of integration of all systems involved in balancing. Therefore, they are a less suitable tool when it comes to understanding which parts of postural control most explain the existence of fall-related concerns. An understanding that could lead to more specific preventative interventions.

The use of a force plate offers more precise ways to look at balance, by measuring center of pressure (CoP) excursions during different tasks. One use of CoP is to look at postural sway, defined by Sheldon as "the constant small deviations from the vertical and their subsequent correction to which all human beings are subject when standing upright" [Sheldon, 1963]. Sway during quiet stance has been used to study postural control and balance [Qiu and Xiong, 2015]. Another way to use CoP, is by exploring its maximum amplitude during a limits of stability (LoS) test in which a person is challenged to lean as far away from their center as they can without losing stability or balance. It is a consistent and reliable measure of dynamic balance [Clark and Rose, 2001]. Conditions that have been known to affect balance – like age, or neurological morbidities – correlate with a reduced LoS [Faraldo-Garcia et al., 2016; Schieppati et al., 1994].

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Both sway and LoS are a result of the different sensory and motor systems working together to create balance. By combining CoP measurements from different tasks with sensorimotor measurements, an explanatory model for the severity of FES-I might emerge.

This new knowledge about how fall-related concerns correlate to different aspects of postural control, could lead us forward in discovering how to improve health and maintain activity levels as a person ages. We therefore aimed to investigate a) which measurements of postural control correlate to falls self-efficacy scores as measured by the FES-I instrument, and b) which sensory and motor systems best explain them.

2. Methods

We carried out a cross sectional study in our movement science laboratory and analyzed predetermined postural control variables in relation to FES-I scores.

2.1. Sample

As this study is a deeper follow-up of an earlier study [Pauelsen et al., 2017], we recruited participants from within that study's sample. That original sample consisted of 153 participants out of 362 randomly selected older adults. Inclusion criteria for the original sample were: community living residents of *(left out for review)* Municipality aged 70 years or older. For the present study the following additional inclusion criteria were used: adequate vision to read 100 pt. large block letters, able to stand unassisted for 30 s or more, able to understand and process simple instructions. Out of 153 from the original sample, 126 were invited to be a part of this follow up study and 45 (36%) of those accepted.

2.2. Data collection

2.2.1. Falls self-efficacy

FES-I data from the previous study was used [Pauelsen et al., 2017]. Items guide the participant to assess how worried they are about falling while carrying out 17 specific activities at home, outdoors and socially. Assessment is done on a Likert scale of 1 (not worried) to 4 (very worried). The sum score ranges from 16 to 64.

2.2.2. Sensorimotor systems

The laboratory visit consisted of an extensive protocol including the following tests:

Bi-ocular vision acuity was screened with the help of an NFD vision chart. This chart is similar to the Snellen chart, but is used at 5 m instead of 6 and is scored using the decimal system instead of the 20/foot system; a score of 1,0 equals 20/20 (normal vision) and 0,5 equals 20/40 (worse vision). Vestibular function was tested with the help of Frenzel glasses. Both passive and active left to right rotation of the neck at different speeds as well as glancing left, right, up, and down was executed during which the occurrence of nystagmus was noted. To measure joint position sense (JPS) in the neck, the participants wore reflective markers in a room with a 3D camera system consisting of 8 cameras (Qualisys, sweden). While sitting with closed eyes, the participant was asked to find the neutral position in their neck. Then, the tester asked the participant to rotate their head to the left to approximately 45° and then reposition to their earlier marked neutral position. This rotating and repositioning was done 6 times per direction (left and right). The absolute error mean of six trials was used.

We assessed pressure sensibility around the ankles by using monofilaments of different stiffness on the lateral malleoli (increments: 0.4, 2, 4, 10, 300 g of linear pressure). Each monofilament – starting with the lightest – was tested 3 times on each malleolus until the participant felt the touch of the filament. JPS in the knee and foot were assessed with the use of a Biodex System 3 machine. The knee repositioning was done at 30° flexion from 90° and the ankle was repositioned to 5° dorsal flexion from 20° plantar flexion. The absolute error mean of three trials was used.

Maximum isometric strength of muscles in lower limb were also measured with the Biodex System 3, which measured the maximum torque in muscles around the hip joint – extension and abduction, the knee joint – flexion and extension, as well as the ankle – dorsal and plantar flexion. Maximum torque of three trials was used.

Participants performed a custom made reaction time test (RT) on the laboratory computer; at random time intervals, a visual and audio cue was produced at which the participant had to push a button as fast as possible. The average of five attempts was used.

2.2.3. Center of pressure (CoP)

Using a Kistler force plate, sampling at 3000 Hz, we measured CoP during quiet stance during four different trials of 30 s each: stable, or hard, surface with eyes open (SEO), hard surface with eyes closed (SEC), unstable, or soft, surface with eyes open (UEO), and soft surface with eyes closed (UEC). Foot placement was standardized by standing with the first metatarsal heads at a distance equal to 75% of the width between the anterior superior iliac spines. Rotational angle of the foot placement was self-chosen. Instructions were to stand up straight, look at the dot on the wall and stand as still as possible. For the eyes closed trials, we instructed the participants to first look at the dot on the wall and then close their eyes when they felt ready. When the participant closed their eyes, the tester made a mark in the measurement. Each trial lasted 30 s. We also used the force plate to measure the participant's limits of stability (LoS) by asking the participant to lean as far as possible in the anterio-posterior (AP) and medio-lateral (ML) directions without moving their feet, nor lifting toes or heels.

2.3. Calculation of outcome variables

We used MATLAB R2017a (MathWorks®, Massachusetts, USA) to generate the CoP trajectories and apply a lowpass butterworth filter with a cut-off at 10 Hz. Then we extracted several classical measures, which describe the participant's sway: AP and ML maximum amplitude, mean velocity, and 95% confidence ellipse of the total CoP signal. The ellipse is based on a principal component analysis of the CoP data points to establish the angle of the ellipse. Then, the smallest ellipse possible is drawn, still including 95% of all data points of the CoP trajectory. We also extracted the maximum AP and ML amplitude measurements for the LoS test.

2.4. Statistical analysis

To calculate the descriptive statistics, SPSS for Windows 24 (SPSS Inc., Chicago, Illinois, USA) was used. After which we used SIMCA 14.0 (Umetrics AB, Umeå, Sweden) to fit a hierarchical orthogonal projection to latent structures regression (O-PLS). O-PLS is a modification of the more classic – principal component based – partial least squares regression (PLS) and we used it to model explanatory abilities of our postural control measurements for FES-I, because the modelling technique allows for many collinear predictors [Eriksson et al., 1999]. The orthogonal modification made during an O-PLS removes the orthogonal – or non-correlated – information from the variability in X. This improves the interpretability of the model. Moreover, O-PLS can handle noisy data structures [Trygg and Wold, 2002]. In this hierarchical model, the base model shows (a) which measurements of postural control correlate to falls self-efficacy scores as measured by the FES-I instrument, and the top model shows (b) which sensory and motor systems best explain those measurements of postural control. In other words, the top model creates a new O-PLS regression with the sensory and motor variables as repressors to explain the model created in the base model (the loadings of that model become the Y values in the top model). The interpretation should be read as which sensory and motor

Table 1
Characteristics and sensorimotor measurements of the participants.

Characteristic	All (n = 45)	Women (n = 28)	Men (n = 17)
FES-I (mean ± SD)	21 ± 4.5	22 ± 4.4	20 ± 4.7
Age (mean ± SD)	75.2 ± 4.5	76.0 ± 5.0	73.9 ± 3.3
Height (mean ± SD)	167.3 ± 9.9	161.8 ± 7.1	179.5 ± 6.7
Short physical performance battery	11 ± 2.2	10 ± 2.4	11 ± 1.7
Reaction time (mean ± SD)	397 ms ± 106 ms	423 ms ± 125 ms	355 ms ± 46 ms
Sensory measures			
Eye sight (corrected, mean ± SD)	0.75 ± 0.18	0.73 ± 0.20	0.79 ± 0.15
Pressure sensitivity at malleoli (median, range)	2 g (0.4–300 g)	2 g (0.4–300 g)	4 g (0.4–300 g)
Positive nystagmus with Frenzel glasses (n)	3	1	2
Joint position sense (mean absolute error and sd)			
Knee, L	6.32° ± 5.62°	7.34° ± 6.57°	4.64° ± 2.82°
Knee, R	5.60° ± 3.76°	5.71° ± 4.22°	5.43° ± 2.87°
Ankle, L	4.61° ± 2.93°	5.30° ± 3.13°	3.52° ± 2.15°
Ankle, R	5.13° ± 3.91°	5.79° ± 4.24°	4.03° ± 2.99°
Neck, L	4.02° ± 3.15°	4.51° ± 3.24°	3.21° ± 2.69°
Neck, R	3.97° ± 2.99°	4.24° ± 3.10°	3.54° ± 2.64°
Strength measures (torque in Nm, mean ± SD)			
Hip extension, L	47.29 ± 19.89	37.67 ± 12.06	63.14 ± 20.09
Hip, extension, R	51.68 ± 22.86	40.24 ± 11.96	70.53 ± 24.02
Hip, abduction, L	51.19 ± 23.73	39.73 ± 16.25	70.06 ± 21.98
Hip, abduction, R	55.53 ± 24.87	43.73 ± 15.02	74.96 ± 25.67
Knee, extension, L	86.61 ± 29.04	71.18 ± 19.33	112.03 ± 24.05
Knee, extension, R	84.25 ± 29.83	69.11 ± 19.97	109.18 ± 26.45
Knee, flexion, L	67.79 ± 24.51	53.94 ± 12.50	90.61 ± 22.29
Knee, flexion, R	70.45 ± 26.87	55.91 ± 15.65	94.39 ± 24.21
Ankle, dorsal flexion, L	21.79 ± 7.91	19.59 ± 6.87	25.42 ± 8.17
Ankle, dorsal flexion, R	23.01 ± 9.01	20.77 ± 5.84	26.70 ± 11.70
Ankle, plantar flexion, L	85.48 ± 35.23	69.32 ± 24.65	112.11 ± 33.84
Ankle, plantar flexion, R	81.79 ± 35.53	67.46 ± 20.16	105.39 ± 35.17

systems correlate to FES-I *through* their correlations postural control measures. This method was chosen because the subsystems cannot and should not be considered to individually influence FES-I as they together, in compensatory and interactive systems, produce balance. All CoP and torque data were normalized to height before analysis.

2.5. Ethical considerations

Written informed consent was acquired from all participants. This study design was approved by the Regional Ethical Review Board in Umeå, Sweden (ref no. 2015-182-31) and is in accordance with the 1964 Helsinki declaration and its later amendments.

3. Results

All participants were included in the analyses and their characteristics as well as their sensory and motor measurements are shown in Table 1. All but a few participants were able to perform 30 s of quiet stance during all testing circumstances (see Table 2).

Fig. 1 illustrates – of one participant – what the CoP trajectories and the 95% confidence ellipses of the four different tests can look like. The figure and Table 2 show how the ellipse surface area becomes larger when the participant closes their eyes and/or stands on a soft surface.

3.1. Base model

When using the variables shown in Table 2 to build an O-PLS model for the explanation and prediction of the FES-I scores, a significant and robust model emerges. The model shows that 39% of the variance in FES-I can be explained through the included variables. Two components are identified – one orthogonal.

The model clearly shows that the trials in which eyesight is eliminated (SEC and UEC), do not contribute much to the model (Fig. 2). They have weak and non-significant loadings. The limits of stability test (stab) and the UEO test are the strongest contributors.

3.2. Top model

When building the top model and adding the sensory motor variables to explain the results of the base model, a significant and robust model emerges that explains the base model to 40%. The loadings are shown in Fig. 3. Besides lower limb strength, we see the following significant variables: reaction time, eye sight, neck proprioception, and knee and ankle proprioception.

The strongest correlations to the base model amongst the lower limb strength variables are found in knee extension and flexion, as well as hip extension (Table 3). The strongest correlation amongst the rest of the systems is shown by reaction time, followed by left sided lower limb proprioception.

4. Discussion

We set out to discover which CoP measurements best describe FES-I variance and in turn, which sensorimotor systems best describe those CoP measurements. We fitted a robust and valid hierarchical O-PLS model showing which variables within those models are significant contributors. The results showed that the CoP variables could explain 39% of the FEZ-I and that limits of stability and the UEO were the strongest postural control variables, as presented in the base model. The result of the top model revealed that 40% of the base model could be explained by the sensory and motor variables, where the significant contributors were lower limb strength, reaction time, eye sight and left sided neck, knee and foot proprioception.

In the base model we found that mainly quiet stance trials with eyes open and limits of stability trials were able to address the variance in FES-I. A larger sway in quiet stance and smaller amplitude in LoS were associated with a higher score on the FES-I. There is very little literature on the relationship between fall-related concerns and postural control in community dwelling older people. Maki demonstrated a larger sway during quiet stance for those with fall-related concerns and discussed the need to investigate whether there is a true physiological decline or if the concern was impacting the postural control [Maki et al., 1991]. In our study, we tested the sensory and motor systems separately and found that declines in many of the systems were strongly correlated with the CoP variables that were associated with an increase in FES-I score. This indicates a true physiological decline, but does not exclude the possibility of fall-related concerns causing compensatory strategies, like co-contraction, as well. Another study made an important distinction between general anxiety and acutely induced fear of falling through postural threat [Sturnieks et al., 2016]. Participants without general anxiety would produce a smaller sway when faced with postural threat, while participants with general anxiety would produce a larger sway when faced with the same challenge. That implies that those who have a more general anxiety would have more difficulty controlling compensatory strategies.

Introducing this number of sensory and motor systems into a model of postural control and falls self-efficacy, has to our knowledge not been attempted before. In a study from 2010, physiological fall risk as a total score was compared with FES-I [Delbaere et al., 2010]. Unfortunately, the physiological score used was not sub-defined, even though it does

Table 2
30 s of quiet standing during 4 different circumstances and maximum limits of stability. Means and standard deviations of the absolute values.

Trial	95% confidence ellipse (mm^2)	Max medio-lateral amplitude (mm)	Max anterio-posterior amplitude (mm)	Mean velocity (mm/sec)
Hard surface, eyes open (SEO) n = 45	326,0 ± 452,5	15,7 ± 12,0	32,0 ± 20,8	19.2 ± 16.4
Hard surface, eyes closed (SEC) n = 43	378,3 ± 439,3	14,7 ± 8,3	40,2 ± 20,6	29.3 ± 26.4
Soft surface, eyes open (UEO) n = 42	1039,2 ± 515,9	29,6 ± 9,3	53,9 ± 16,4	36.3 ± 11.6
Soft surface, eyes closed (UEC) n = 41	2590,0 ± 1218,6	50,3 ± 39,1	93,4 ± 25,1	76.4 ± 34.6
Limits of stability n = 43	–	209,8 ± 60,8	142,9 ± 28,0	–

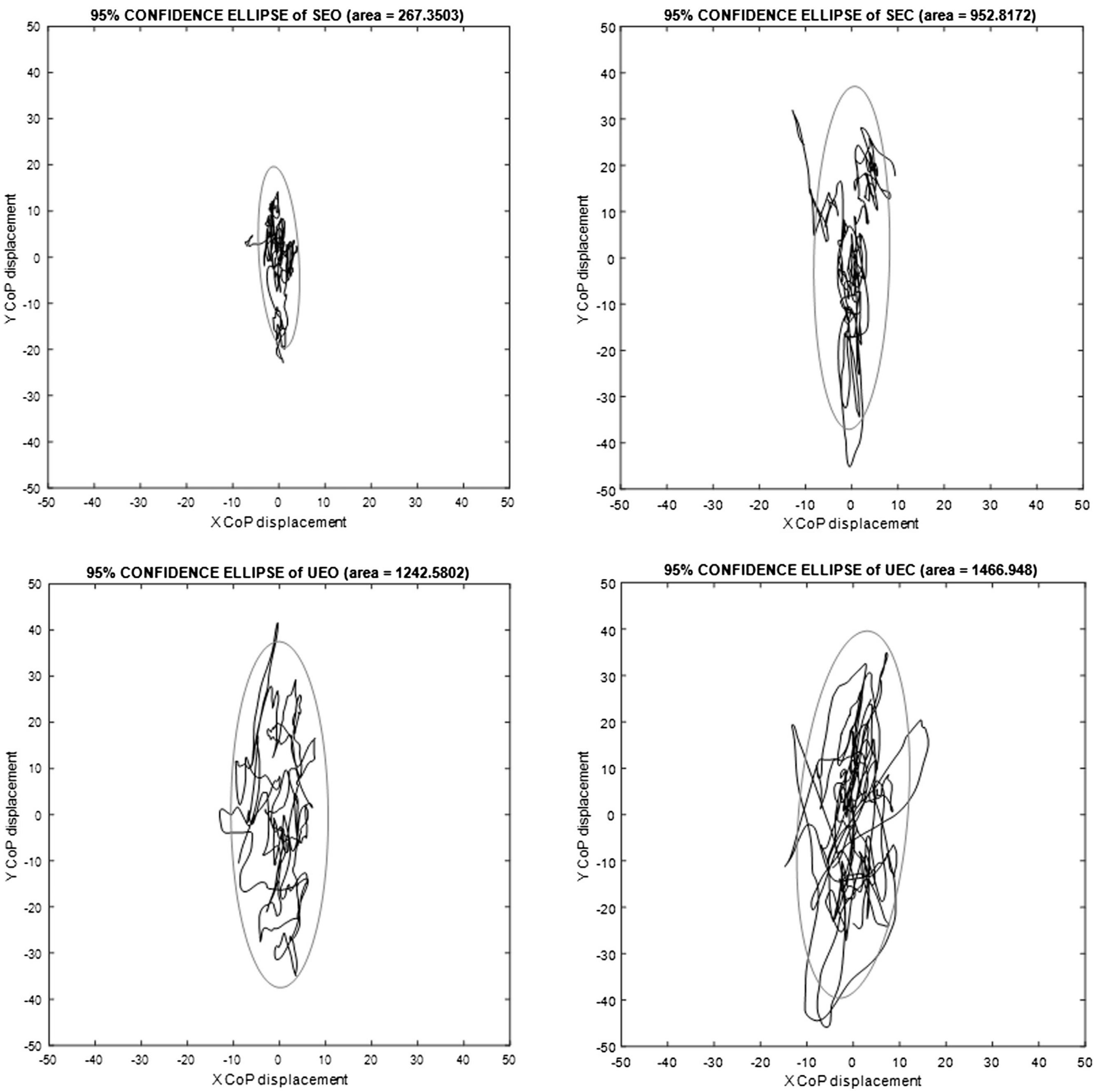

Fig. 1. A visual representation of the CoP trajectory during the 4 different tests. The participant is a 72 year old woman with above average errors on lower limb proprioception and touch sensation. Her FES-I score was 20.

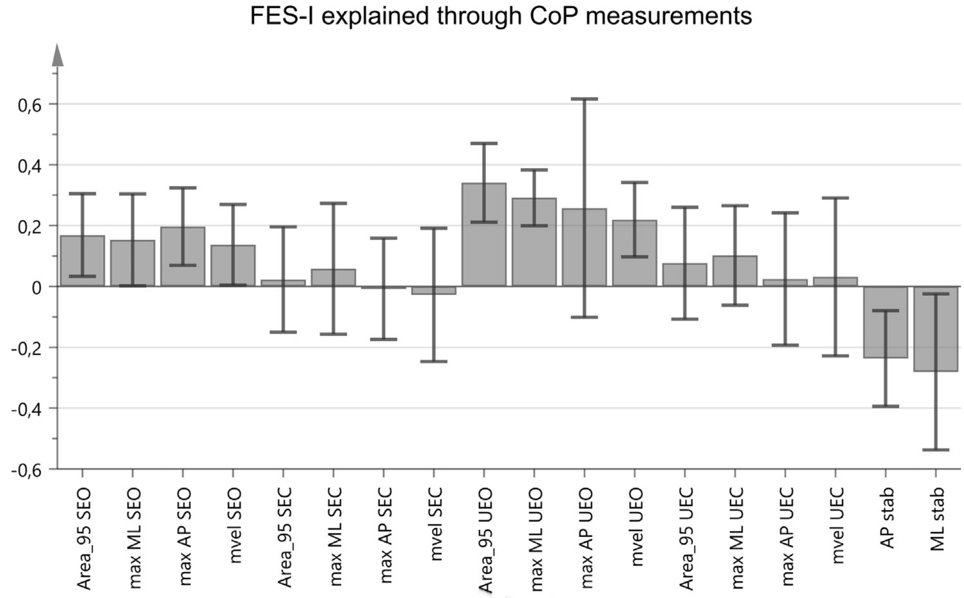

Fig. 2. Loading plot for the base model of FES-I. Error bars not including 0 indicate significant loadings. Variables have been centered and scaled for unit variance.

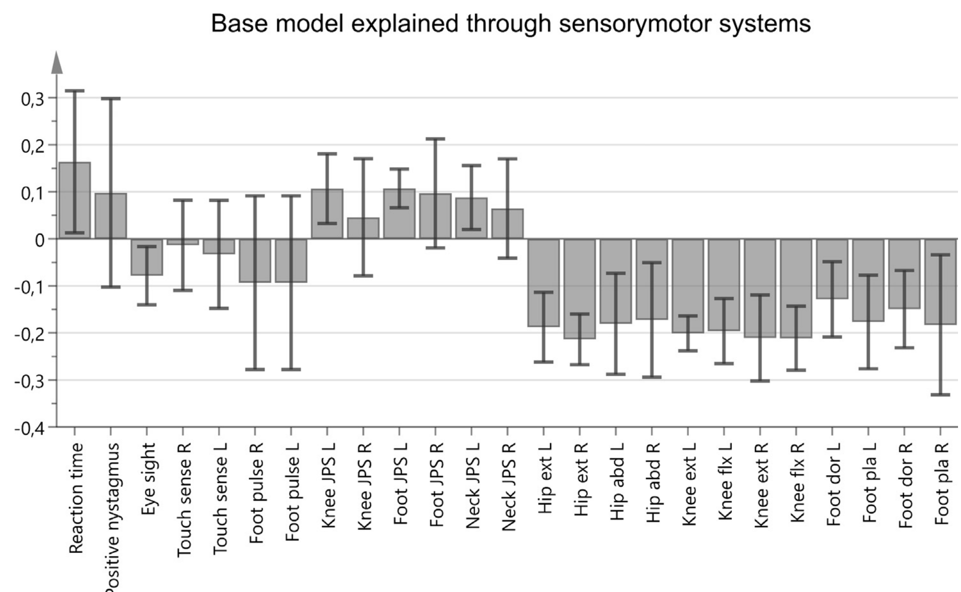

Fig. 3. Loading plot for the top model of FES-I. Error bars not including 0 indicate significant loadings. Variables have been centered and scaled for unit variance.

consist of five separate sensorimotor measures. The aim of that study was rather to determine the difference between physiological fall risk and FES-I than to see how the two could be mediating each other. They saw in their follow up that both aspects separately correlated to fall risk, further validating the need to research mechanisms behind the variance of FES-I.

Our results of the top model show that all measured sensory and motor systems – apart from the vestibular system – are involved in controlling CoP, but all with different levels of importance. Two important things stand out. The first is that proprioception only shows significant loadings to the left side of the body. Since we did not systematically start the assessments with the right side, learning is not a factor in this. We measured the neck and the leg in two different systems (3D cameras and Biodex respectively), but the neck and the leg both show this bias towards the left side. Group error averages do not show significant differences between left and right and the variances are similar in size. Left or right foot dominance did not show a correlation to the measured proprioceptive errors. It has been shown that left and right proprioception in the legs are individually weighted during perturbation in quiet stance and loss of proprioceptive information from one leg (due to perturbation) is countered by information from the vestibular system rather than the other leg [Pasma et al., 2012]. This supports our findings of different loadings for left and right and suggests that findings of one side may not generalize to the other. However, as we perturbed both sides equally during the standing trials, we have no answer as to why left sided proprioception seems more

Table 3
List of variables with significant loadings in the top model and their correlation coefficients to the base model.

Sensorimotor variable	r
Reaction time	0,538775
Eye sight	−0,287064
Knee JPS L	0,467836
Foot JPS L	0,37937
Neck JPS L	0,318744
Hip ext L	−0,755609
Hip ext R	−0,845208
Hip abd L	−0,769482
Hip abd R	−0,736908
Knee ext L	−0,828598
Knee flx L	−0,837361
Knee ext R	−0,791346
Knee flx R	−0,836008
Foot dor L	−0,545189
Foot pla L	−0,787085
Foot dor R	−0,565373
Foot pla R	−0,830819

important than right. More research is needed to address proprioceptive laterality. The second important thing that stands out is that the vestibular system does not show up in the model. Only three out of 45 participants showed a positive vestibular result and one of those had to be excluded from the modelling due to missing data. The fact that the vestibular system did not show up in our model should therefore not be interpreted as definitive. We deem it of much importance to investigate these aspects further.

When measuring sensory and motor systems, one has to choose between many different available methods. We measured vision acuity, the vestibular system, and touch sensation by use of screening methods, which resulted in data scaled on only a few increments – vestibular only on a dichotomy for nystagmus. While our measurements were accurate, they may not have been precise enough to identify all issues present in our sample. This is reflected in the statistical results, meaning that while eyesight shows a lower loading than the different strengths, this should not be interpreted to mean that eyesight is of lesser importance. However, all but the vestibular system did show up as important within the model. More precise measurements might have given them even more importance.

This sub-sample of 45 participants had one point more on SPPB and one point less on FES-I than the larger sample of 153 used in the previous study, indicating that these participants were slightly fitter and slightly less concerned about falling than the population. This is an inevitable recruitment bias as our testing protocol did require participants to be able to stand and balance without help. If anything, with the strong correlations found between postural control aspects and FES-I, a less fit sample who are more afraid would be expected to make the models even stronger.

Unfortunately, the sample was not large enough to be able to model FES-I and postural control for each sex separately. We have seen that the larger model for FES-I from the previous study, which includes physical performance for both sexes, contains more physical aspects for men than for women [Pauelsen et al., 2017], which suggests that the models of the current study might look slightly different for each sex as well. This should be investigated with a larger sample.

5. Conclusion

As expected, lower limb strength plays an important role in postural control changes correlated to falls self-efficacy, but reaction time, sight, touch sensation, and proprioception are also involved and should not be overlooked. Aspects within sensory input, central integration, and motor output all are correlated to changes in FES-I.

Conflict of interest

There are no conflicts of interest for any of the (co)authors.

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