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## Effect of textured foot orthotics on static and dynamic postural stability in middle-aged females

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### Abstract

Foot orthotics (FO) may be prescribed for a range of lower limb and foot conditions. Prior studies report use of FO in enhancing postural stability in healthy younger adults, and do not control for footwear type. Currently, interest in the effects of FO on postural stability in older adults has increased. Limited reports exist of the effects on postural stability of FO made of combinations of materials, thicknesses and surface textures. In this study 40 healthy females ( $51.1 \pm 5.8$  years) recruited into a within subject test–retest randomised clinical trial were provided with identical footwear and randomised into four FO conditions (control, grid, dimple and plain,  $n = 10$  for each condition). Participants wore the footwear for 4 weeks, a minimum of 6 h/day. A Kistler force plate was used to determine postural stability variables (anterior–posterior displacements and medial–lateral displacements) for each participant in a static position, with eyes open and eyes closed. Base of support was evaluated using the GAITRite<sup>®</sup> system. Each outcome measure was measured at baseline and 4 weeks. Postural stability variables demonstrated no significant differences between the four FO conditions. No significant differences were observed with base of support between the four conditions. We have demonstrated no detrimental effects on postural stability in older females after 4 weeks. This is regardless of orthotic texture and is independent of footwear. Biomechanical or sensory effects of FO on postural stability are still to be determined. These may be dependent on the geometry and texture of the orthotic.

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### 1. Introduction

When walking, the foot is the first point of contact between the body and the external environment, providing sensory information to the central nervous system for stability and locomotion. Postural stability and balance are dependent on the position of the centre of mass of the body and its displacement within the base of support [1]. External disturbance of balance in the anterior–posterior direction is corrected by a synergy of muscle activations known as the hip–ankle strategy [1]. Medial–lateral

disturbances in quiet standing make use of hip muscle activation to maintain postural control [1]. The maintenance of postural stability is also dependent on a range of somatosensory inputs. Tactile sensitivity within the foot has a strong influence on maintenance of postural stability, as evidenced when this sensory input is lost in diabetic neuropathy [2,3]. Vision has a definitive role in postural control [2,4]. Trials of postural control of older subjects who have visual field temporarily removed demonstrate a worse performance of the task than under full vision conditions [4]. A reduction in somatic sensation and vestibular function have been strongly linked to postural instability, to the extent of adversely impacting on balance and contributing to falls in the elderly [4]. The link between postural stability and falls is well documented with postural instability deficits being significant predictors of falls in older adults [5,6].

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Footwear may influence the quality of sensory feedback from the feet and may act as a sensory filter between the feet and the external environment [7–9]. Different types of footwear have been shown to affect balance [10,11], with low-heeled walking shoes giving better control over postural sway and gait performance [7]. Shoe inserts and foot orthotics (FO) are thought to affect sensory feedback from the feet and aid postural control [12]. It has been suggested that soft materials negatively affect stability during locomotion and for optimal stability, shoes with thin, hard FO are preferable [13,14]. Furthermore studies investigating stimulation of the soles of the feet through vibrating FO [15], magnetic FO [16] and textured FO [9,17] are associated with improvements in balance control and reduce postural sway.

Research suggests that older adults may use larger excursions of postural sway to increase information about postural stability from their sensory systems [18,19]. Moreover, falls occur primarily during activity. Walking is essentially an unstable condition with the body's centre of gravity projecting outside the base of support for most of the gait cycle [20]. The breadth of the base of support is critical to stability. Older adults often adopt a wider base of support for this reason. Width of base of support is, therefore, often used as a measure of balance [21].

The purposes of the study were to develop a standardised methodology and evaluate the clinical effectiveness of four differently textured FO on static and dynamic balance variables in an older female population.

## 2. Methods

### 2.1. Subjects

The study design was a prospective pilot single blind randomised clinical trial over 4 weeks. A convenience sample of 40 healthy female subjects was recruited from the staff and student population of the University of Teesside (age =  $51.1 \pm 5.8$  years; height  $1.63 \pm 0.47$  m; weight of  $75.2 \pm 18.9$  kg). All subjects were free of musculoskeletal or neurological conditions that could have affected gait and postural stability. The University School of Health & Social Care Ethics Committee gave approval for the study and informed consent was obtained from all subjects prior to their participation. Fig. 1 shows the progress of subjects through the study.

### 2.2. Footwear

All subjects were fitted with standardised footwear, consisting of a flat, casual, lace-up shoe of appropriate size ('Wash', Hotter (UK) Ltd., Skelmersdale, United Kingdom). An independent observer used randomisation tables to assign 10 subjects per group to their FO condition at baseline. The control group received shoes that were fitted with the standard Hotter shoe insole (Shore Value A20). Subjects in the three intervention groups received experimental FO manufactured in 3 mm EVA (Shore value A50), (Algeos (UK) Ltd., Liverpool, United Kingdom) with differently textured surfaces. The first intervention group ( $n = 10$ ) received shoes fitted

with a flat, plain, and smooth surfaced FO, made of three millimetre thick EVA. The shoes of the second intervention group ( $n = 10$ ) were fitted with a dimpled surfaced FO (1 mm raised circles, with a diameter of 3 mm spaced 5 mm apart covering the entire surface of the FO). The third intervention group had an FO with a raised grid pattern fitted into their shoes (1 mm raised square pyramid shapes, side length two point 5 mm, peaks spaced two point 5 mm apart covering the entire surface of the FO). Fig. 2 shows cross-sections of the dimple, grid and plain FO. The subjects were required to wear the shoes for 4 weeks. The subjects were asked to wear the shoes as their main shoe, for a minimum period of 6 h/day. The shoes had to be worn with the FO fitted within them.

### 2.3. Postural stability and base of support

Postural stability was assessed on a Kistler force platform (Model 9286AA, Kistler, Alton, United Kingdom) by recording the differences between the maximum and minimum ranges of centre of pressure (COP) in the medial–lateral and anterior–posterior directions [13,22,23]. Subjects stood on the force platform wearing the shoes and FO. Each subject stood directly on the surface of the force platform with the feet placed either side of a T-shaped separator. This was done because foot placement is believed to influence postural stability [24]. Each subject's heels were in a standard position 12 cm apart. According to Le Clair and Riach [22] the optimum test–retest reliability is obtained from the Kistler force platform during a period 20–30 s into the recording procedure. The first 10 s of measurements were discarded and the measurements for this study were taken for a period of 30 s at a sampling rate of 50 Hz for each trial [22,23]. Limits of stability recordings were taken in two visual conditions. These were eyes open [EO] and eyes closed [EC]. In EO condition each subject was asked to focus on a 2 cm diameter white spot positioned on a screen 2 metres in front of them at their eye level. This was to standardise the data collection and minimise any variations in postural stability brought about by changes in visual field. All recordings were taken in a quiet laboratory, with a minimum of external distractions.

Base of support was evaluated during walking using the GAITRite system (12 Foot Gold, Electro-Medical Supplies (Greenham) Ltd., Wantage, United Kingdom). This is a portable gait analysis tool consisting of a carpet 460 cm long with an active sensor area of 366 cm long and 61 cm wide. Base of support was defined as the perpendicular distance from the heel point of one footfall to the line of progression of the opposite foot. This is thought to be an important variable in maintenance of stability [21]. Subjects were instructed to walk over the carpet at a normal comfortable walking speed. The first few and last few steps of each trial were discounted to eliminate gait changes associated with acceleration and deceleration, respectively, and to ensure subjects were walking at their preferred walking speed. In order to exclude the first and last few steps of each trial, subjects began walking from a point 3 m in front of the mat and stopped at a point 3 m beyond the end of the mat. Subjects were asked to focus on a cross of 20 cm axis positioned at their eye level at the far end of the walkway. This was to ensure that the GAITRite walkway did not influence the positioning of subject's feet, and allow natural patterns of gait to be recorded. Three trials were recorded for each subject and the mean used for subsequent analysis. Triplicate readings have been demonstrated to have fair to good ICC with respect to test–retest reliability for base of support in older adults [21].

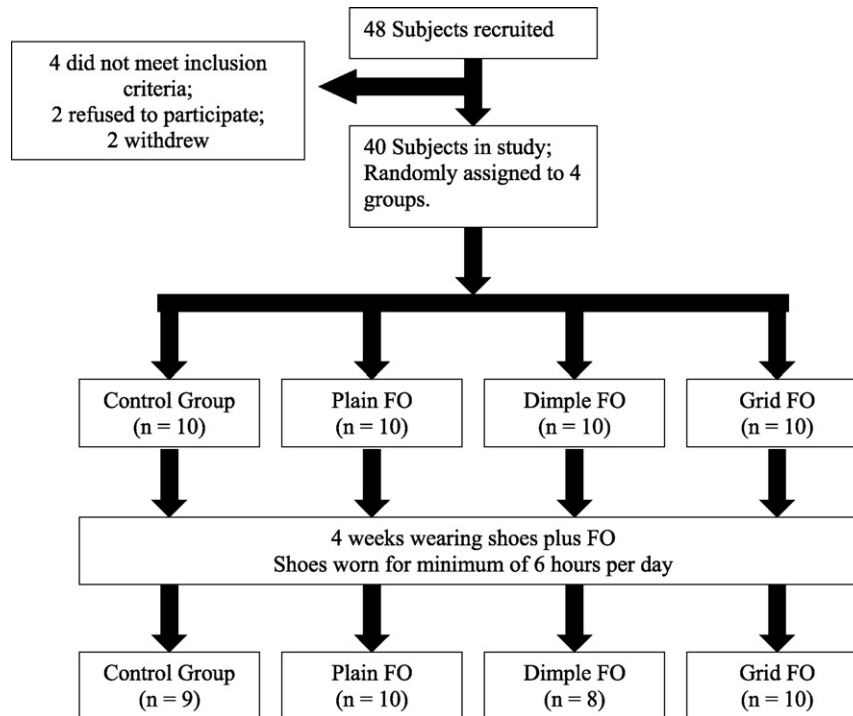


Fig. 1. Flow diagram of progress of subjects through phases of the trial.

#### 2.4. Data analysis

Baseline descriptive information was obtained from each subject. For analysis of static postural stability a  $2 \times 4$  repeated measures ANOVA was carried out on the changes in postural stability over a 4-week period. This was conducted by two independent variables, the four types of FO and the two visual conditions (eyes open and eyes closed). For analysis of dynamic postural stability a repeated measures ANOVA was also undertaken. Data

analysis was carried out using SPSS for Windows (SPSS Version 11.5, SPSS UK, Woking, United Kingdom) was used to perform these analyses and all data were assessed at the 5% level of significance. Analysis of the results aimed to determine whether there was:

- (i) Any significant change in the dependent variables of the range of COP for medial–lateral and anterior–posterior stability during quiet standing between the four FO groups over the 4-week experimental period. This was assessed by looking at the change in the postural stability variable over time for the conditions eyes open and eyes closed.
- (ii) Any increase or decrease in the dependent variable of base of support during walking between groups over the 4-week experimental period for each group. This was also done by using the change in the outcome variable.

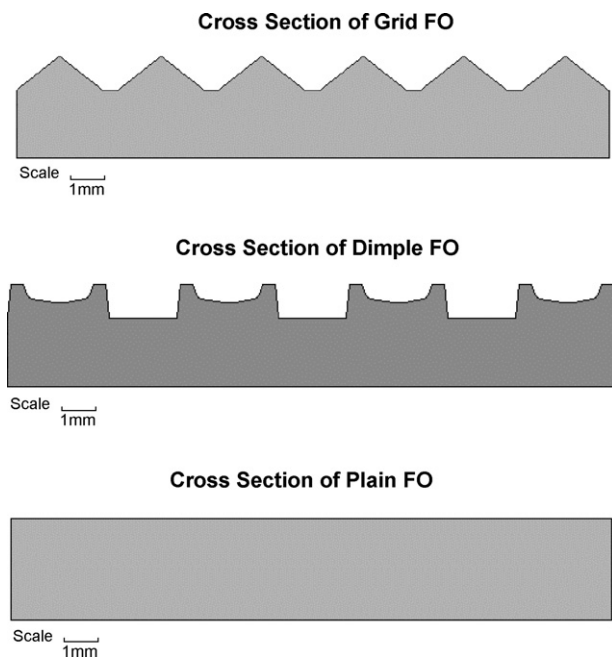


Fig. 2. Cross-section of the grid, dimple and plain foot orthoses (FO).

### 3. Results

Three subjects withdrew from the study during the 4-week period (one from the control group and two from the dimple FO group). Data from 37 subjects were available for final analysis (Fig. 1).

There were no significant differences between the physical characteristics of the groups at baseline for age ( $p = 0.894$ ), height ( $p = 0.267$ ) and weight ( $p = 0.782$ ).

#### 3.1. Medial–lateral postural stability

At baseline, no statistically significant differences were observed between the FO groups in the EO ( $F = 0.411$ ,

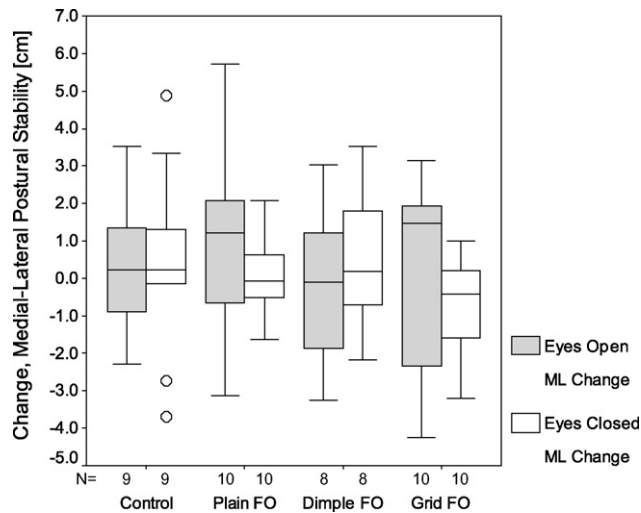


Fig. 3. Box and whisker plots of median values of the change in medial-lateral postural stability for eyes open (shaded boxes) and eyes closed (open boxes). Bars represent the range of values. Open circles represent outliers. No statistically significant differences were observed for eyes open or eyes closed conditions for any of the FO. There was no significant interaction between visual condition and FO condition.

$p = 0.746$ ) or EC conditions ( $F = 0.479$ ,  $p = 0.699$ ) for medial-lateral postural stability. Over 4 weeks, there was no significant difference in the change in medial-lateral postural stability when considering the EO and EC conditions ( $F = 0.273$ ,  $p = 0.605$ ). Also in this time, there was no significant difference in the change in medial-lateral postural stability when considering the FO conditions ( $F = 0.677$ ,  $p = 0.572$ ). There was no significant difference in the interaction of visual conditions and FO conditions ( $F = 0.590$ ,  $p = 0.626$ ). Box and whisker plots of the change

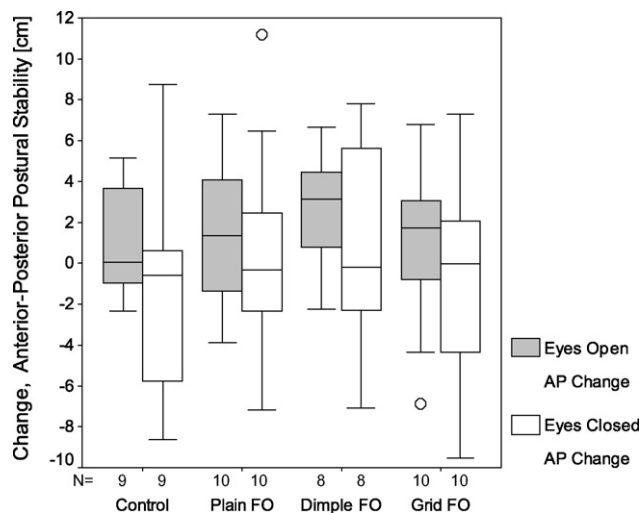


Fig. 4. Box and whisker plots of median values of the change in anterior-posterior postural stability for eyes open (shaded boxes) and eyes closed (open boxes). Bars represent the range of values. Open circles represent outliers. No statistically significant differences were observed for eyes open or eyes closed conditions for any of the FO. There was no significant interaction between visual condition and FO condition.

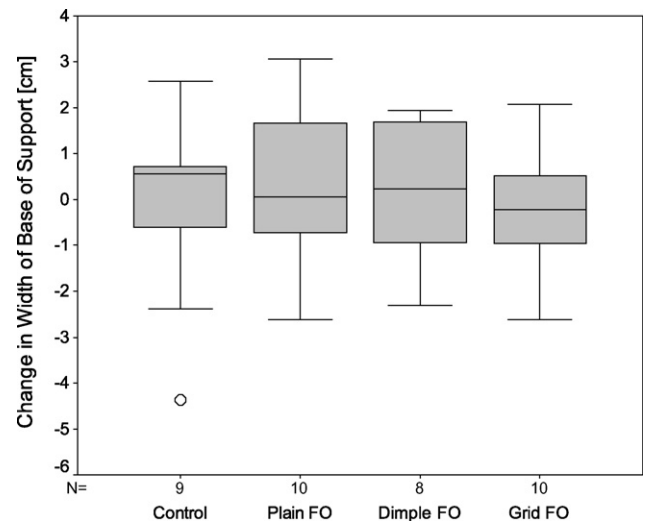


Fig. 5. Box and whisker plots of median values of the change in base of support for each condition. Bars represent the range of values. The open circle represents an outlier. No statistically significant differences were observed between FO conditions.

in medial-lateral postural stability for EO and EC conditions for each FO are shown in Fig. 3.

### 3.2. Anterior-posterior postural stability

At baseline, no statistically significant differences were observed between the FO groups in the EO ( $F = 0.475$ ,  $p = 0.702$ ) or EC conditions ( $F = 0.240$ ,  $p = 0.868$ ) for anterior-posterior postural stability. Over 4 weeks, there was no significant difference in the change in anterior-posterior postural stability when considering the EO and EC conditions ( $F = 3.029$ ,  $p = 0.091$ ). Also in this time, there was no significant difference in the change in anterior-posterior postural stability when considering the FO conditions ( $F = 0.567$ ,  $p = 0.641$ ). There was no significant difference in the interaction of visual conditions and FO conditions ( $F = 0.213$ ,  $p = 0.887$ ). Box and whisker plots of the change in anterior-posterior postural stability for EO and EC conditions for each FO are shown in Fig. 4.

### 3.3. Base of support

At baseline, no statistically significant differences were observed between the FO groups ( $F = 0.479$ ,  $p = 0.699$ ) for base of support. There was no significant difference in base of support ( $F = 0.019$ ,  $p = 0.892$ ) over the 4 weeks of the study. Also in this time, there was no significant difference in the base of support when considering the FO conditions ( $F = 1.481$ ,  $p = 0.238$ ). There was no significant difference in the interaction of base of support and FO conditions ( $F = 0.265$ ,  $p = 0.850$ ). Box and whisker plots of the change in base of support for each FO are shown in Fig. 5.

#### 4. Discussion

We have developed a methodology that controlled for shoe type and foot position, and that has determined the effect of different FO profiles on postural stability. We have assessed this for both static and dynamic postural stability. This study demonstrates that when shoe type is standardised there is no detrimental effect on postural stability over a 4-week period in middle-aged women using FO with differently textured surfaces. This is our finding in both static and dynamic conditions, for subjects with eyes open or eyes closed. These data are in agreement with that of Percy and Menz [25]. These workers found no improvement in postural stability in footballers (soccer players) using FO. However, these workers used trunk displacements as their outcome measure, not a balance platform as was used in this study. Additionally, they studied young healthy adult males.

Previous FO studies have demonstrated that footwear type and FO materials can influence postural stability in young adult males [8,17,26]. However, we believe this to be the first study that controls the footwear type along with the FO in an older adult female population. Mündermann et al. [27] have investigated the kinematic effects of FO in a standardised running sandal. However, this study only considered footwear worn when taking part in an exercise activity and not the main shoe worn by the individual for prolonged time periods. In addition, subjects only wore the shoes for running purposes for a 2-week period unlike in the current study which was for continuous 4 week period as the main shoe. The only other study where individuals wore FO for a prolonged period of time was carried out in young adults, and only investigated comfort perception and reported injury rates [8].

The Shore values of the FO and the integral insole in the shoes in this study were relatively similar. This ensures that the variability of inserts is reduced, increasing the validity of the measurements obtained. Thin flat shoes allow for better somatosensory feedback [7,6]. In older women, their ability to perform simple functional tasks was demonstrated to be significantly better when wearing what Arnadottir and Mercer [7] termed walking shoes. In our study no detrimental effects of the thin FO were observed in static or dynamic conditions. Previous work by other workers on the effect of shoe sole properties and textured FO on ankle movement discrimination in younger age-groups demonstrate improvements in postural stability [9,13,28]. For example, the use of textured inserts in athletic shoes was demonstrated to improve significantly the detection of ankle movement in young healthy females, relative to those who had no inserts [9]. In a subsequent study, the ability of young female soccer players to discriminate extent of ankle inversions was improved when using textured FO [28]. Both these studies assessed ability to detect and correct disturbances in joint angles at the ankle, not maintain static postural stability. It may be that the mechanisms used to make these successful value judgements are different to

those used in maintaining static postural control. It should also be noted that in these previous studies participants were required to undertake balance training programmes in addition to using the FO, which may have positively influenced the results unlike in the current study where no balance training took place.

In a study on the effect of vibrating insoles on balance control Priplata et al. [15] based their findings on head–arm–trunk displacement measured by the Vicon camera system. Displacement data were acquired from a single marker attached to the right shoulder of each participant. They reported improvements in postural sway in older adult subjects while the subjects were positioned to stand on vibrating FO for short intervals of time. In the current study, subjects wore the FO over a period of weeks within their shoes, which we believe should give a more representative demonstration of the efficacy of the intervention. This is particularly so, as one would expect FO to be worn in shoes on a daily basis to prevent falls. We also believe that in using the Kistler force platform to record the COP, we have truly recorded whole body postural stability adjustments. The Kistler force platform has previously been shown to be able to detect and discriminate COP deviations in older adults presented with a range of challenges to postural stability [23]. COP has been shown to be influenced by height and foot positioning [24]. In this current work, we found no significant differences in the heights of the women and standardised their foot position to reduce the influence of these variables on COP. Postural stability in the current study was assessed by recording the range of anterior–posterior and medial–lateral displacements. It is believed that different mechanisms of postural control are used to correct anterior–posterior deviations from those used to correct medial–lateral deviations [1]. By assessing each separately we believe we have explored the possibility that the wearing of textured FO could have influenced postural stability in one direction more than in the other. This does not appear to have been the case for any of the FO.

Olmstead and Hertel [11] suggest that foot type may also influence the effect of the orthotic employed, and for future work this may need to be detailed in the inclusion/exclusion criteria. Previous work by Rome and Brown [14] demonstrated improvements in postural stability in younger adults who were diagnosed with excessive foot pronation when a rigid FO was used. It was suggested that rigid FO stabilised the rear foot of the subjects in the study, reducing internal tibial rotation. Previous work by other groups has suggested that rigid FO provide greater stimulation of mechanoreceptors in the feet [13]. A study of military personnel [8] observed a reduction in the number of reported injuries during the study in subjects using a variety of different FO of varying degrees of hardness, viscosity and elasticity.

Another important factor is the geometry of the insole. With the exception of the control insole, which was slightly contoured, the issued insoles were flat in profile. Insoles moulded to the geometry of the foot may well have a

different effect, although flat FO have been used to positive effect in other studies of younger adults [28]. We would therefore postulate that in older adults there is a requirement for the FO to be contoured in order to improve postural stability.

There is a possibility that older individuals use different postural control strategies in order to maintain centre of gravity over the base of support [4,19,30]. This may explain why we did not observe the expected reduction in excursions of centre of pressure over the 4 weeks. Older adults have been shown to have increased muscle activity in the lower limb [29]. This could arise as a consequence of loss of somatosensory processing leading to reliance on motor strategies to maintain posture. Morasso and Sanguineti [30] propose that increased muscle stiffness is used as inadequate compensation for a reduction in sensory inputs. This could be of particular relevance to conditions such as diabetic neuropathy. Mackey and Rabinovitch [19] observed that large excursions of postural sway in quiet standing were associated with a better ability to generate ankle torques necessary to maintain upright posture in older women. It is possible that the neuromuscular mechanisms involved in maintaining and controlling stable quiet standing are different to those mechanisms we use to react appropriately to disturbances of balance [1,19,30]. Providing increased sensory inputs via grid textured FO may be an inexpensive and effective way of reducing the risk of falls in this client group. Sensory deficit in the feet and its effect on balance is another area for potential study. No sensory screening of subjects was carried out in this study, but such screening may reveal more information on the relationship between postural sway and footwear/insoles.

Standardisation of the foot position for all subjects increases the reliability of the measurements. It has been well documented that alteration of foot width in quiet standing may influence postural stability [24]. All subjects were required to align their foot position with a specially adapted T-bar shape. In general, no other studies investigating postural stability have standardised foot position. Anthropometric differences such as height have been demonstrated to influence the performance of balance tasks [24]. For each of the FO groups in this study, there were no statistical differences for age, weight or height. We are therefore confident that our COP measurements are valid.

We recognise that we have some limitations to our study. Our findings may have limited application to an older, frailer female population, as we sampled a relatively healthy middle-aged female population recruited from the staff and student community. There is therefore the possibility of sample bias. Each of the FO groups contained only 10 subjects. This raises the possibility of type II sampling errors, and erroneous acceptance of the null hypothesis. We would however argue that this study was a much a pilot study in testing the methodology, and that many previous studies claiming improvements in postural stability have based conclusions around small convenience sample sizes

[9,15,16,27,28]. Long-term wear of the FO could have influenced the functionality of the devices. There is a possibility of changes in the material properties of the EVA used in the FO during daily wear. For example, the materials could have become softer. We did not assess property changes to the FO in this study, but we feel it would be safe to assume few changes would occur that could influence the results of this study. The materials we used are routinely employed in the construction of FO. Property changes to the FO materials could be assessed in future work.

In conclusion, in static balance and in our 30-m walk, there were no decrements or improvements in postural or walking stability over a 4-week period of continuously wearing any of the shoes with any of the FO. The FO condition had required the shoe's original insert to be removed, and the thin FO placed inside the shoe. This in itself could have enhanced the sensory feedback to the feet, although no significant effect on the static and dynamic balance was observed. For future work we propose that suitably adapted and contoured inserts be used to reflect a more realistic control condition.

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