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Postural Stability Analysis of Center of Pressure Data from Normal and Flat Feet during Upright Standing using Signal Energy

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This study applied the signal energy (SE) method using center of pressure (COP) data to quantify postural stability. A randomized control trial was conducted to determine differences between subjects with normal and flat feet. Fifty-four subjects aged 18–30 years participated; 37 normal and 17 flat foot subjects. All subjects undertook tasks commonly used to quantify postural stability. Measurements quantified using the SE method were compared with those quantified using commonly employed methods for assessing postural stability. Using the SE method, total energy values required for maintaining postural stability with flat feet differed significantly ($p < 0.05$) compared with normal feet when eyes were open and closed in the medial-lateral direction. Signal amplification and observed variance were demonstrated by the SE method, and can be used to clarify differences in quantitative postural stability between normal and flat feet. Hence, SE might be a valid biometric method for balance control assessment; however, further study of subjects from different age groups is required to validate the application of the SE method in postural stability quantification.

Keywords: Flatfoot, postural stability, signal energy, information systems, physical sciences and engineering

Foot arches in human feet absorb shock and are vital in maintaining balance. The contour of the foot widens with age, and inadequate exercise and increased body weight may cause foot muscle deterioration and increase flat footed structures [1-4]. Foot structure tends to worsen as the shock-absorbing effect deteriorates and as plantar pressure distribution abnormalities increase. An abnormal arch structure may cause plantar fasciitis, plantar pain, tendonitis, foot pain, muscle ache, knee pain, back pain, or other problems. Furthermore, the ability of people with flat feet to maintain postural stability deteriorates and leads to an increased risk of falling [5, 6]. Therefore, strategies such as corrective aids to control abnormal foot gait biomechanics are used to alleviate the discomfort experienced [7]. However, appropriate assessment instruments that quantify changes in postural stability are still required [8, 9].

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A cheap, easy to use, and portable measurement system, such as a force plate with the appropriate software used to evaluate postural stability differences, is required for routine measurements in the field [10, 11]. Time-series analysis based on center of pressure (COP) data is mostly used to examine postural stability while standing upright on a force plate, and can be integrated with a low cost, portable measuring device [10]. Traditional COP-based methods use raw data means to quantify postural stability [12]. The multiscale entropy (MSE) method is used to measure the complexity index (CI) to quantify postural stability, and is considered to be a better measurement for analyzing human non-stationary fluctuations in posture sway data [13]. For this, the signals are decomposed by the time-frequency Empirical Mode Decomposition (EMD) method into intrinsic mode functions (IMFs) [14]. Moreover, entropy methods need to explore different parameters and combinations of IMFs [15-17]. In this study, we used the signal energy (SE) method to evaluate posture stability from COP signals, which has not been used as a COP-based parameter in this field to date.

Previously, energy cost measurements have been used to assess differences in energy consumption between people with normal feet, flat feet, and those using foot arch supports [4,18,19]; however, implementing such measurements during daily life activities would be time consuming and complicated. The SE method uses digital signals that are collected during routine activities, which can be converted into energy values for analysis [20]. The converted SE signals can be applied to different fields, such as flight [21], radio signal distribution [22], voice recognition [23], and medical engineering in ways that include improving the ECG signal approximation method [24], estimating survey flight time using SE [25], and enhancing magnetic resonance force microscopy detection [26]. This approach is different from the concept of energy expenditure through general movement, which involves calorie or oxygen consumption [4, 18]. The SE signal is calculated according to the magnitude of the signal amplitude and observed variance, and data are analyzed by calculating the squared amplitude of the signals; this method is suitable for analyzing data with subtle changes [26].

In this way, the structure of the foot arch is important for posture [1,5,27], and is commonly used for assessment during physical examinations [28] and in COP signal analysis [12]. Further studies have

evaluated foot structure, risks to postural control with abnormal foot structures [6,29], and age-related degeneration of the foot arch [2,30]. Therefore sensitive and appropriate assessment instruments are required to quantify subtle changes in postural stability. The aim of the present study was to further investigate the effectiveness of incorporating the SE method in postural stability measurements from the gait function test and other commonly used parameters during quiet standing tasks to determine differences between subjects with normal and flat feetThe aim of the present study was to further investigate the effectiveness of incorporating the SE method in postural stability measurements from the gait function test and other commonly used parameters during quiet standing tasks to determine differences between subjects with normal and flat feet.

METHODOLOGY

-Subjects and Equipment

Fifty-four healthy young adults aged from 20 to 30 years (mean age: 23.28 ± 2.10 years) participated in this study. The inclusion criteria were: no history of mental illness, or musculoskeletal or foot disorders, and no foot surgery or other invasive foot treatment procedures. Before participating in the experiment, all subjects provided written informed consent (Research Ethics Committee: NTU-REC No. 201206HS011) and relevant personal information. Subjects were then divided into two groups, the flat foot and the normal foot groups, based on static ink footprints while standing in a relaxed position. The footprints were classified using the footprint classification method [2], whereby the arch index ratio R was calculated; R ≧ 1 was classified as a flat foot (Figure 1). There were 17 subjects (mean age: 23.62 ± 2.29 years) in flat foot group and 37 (mean age: 22.53 ± 1.42 years) in the normal foot group.

In this study, COP movement trajectory data was collected using a force plate (OR6-7-2000; AMTI, Watertown, MA, USE; 50.8 × 46.4 cm). The analyzed COP sway time-series in the anterior (AP) and medial-lateral (ML) directions were derived from signals that were continuously recorded for 60 s at a sampling rate of 100 Hz; the obtained time-series had 6000 sample length. The data were used to investigate the quantitative postural stability of the subjects when standing upright in each trial.

-Method

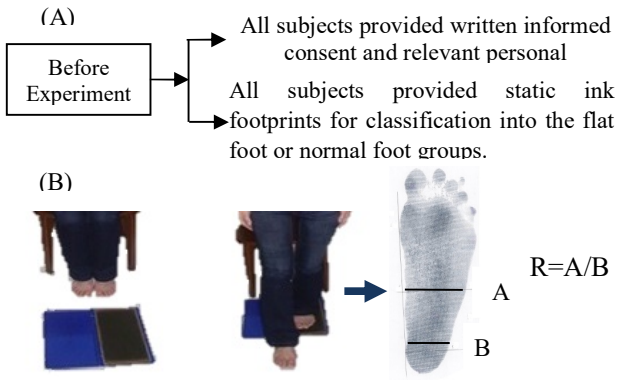

Figure 1. Pre-testing procedures. (A) All subjects provided written informed consent and footprints before the experiment. (B) The arch index ratio R of the footprint was collected from a step on a sheet of paper then calculated by dividing the width of the center of the footprint A by that of the posterior region B

Gait Function Test

This study involved three commonly-used gait function test items used for balance measurement in relevant clinical research: the 6-min walk (6 MW) [31, 32], timed up and go (TUG) [33], and one-leg standing (OLS) test with eyes closed (EC) [34]. These measurements have been used for the functional assessment of lower limb muscle strength in recent years. The literature found that lower limb muscle strength decline and related poor physical posture performance affects daily physical activity function and balance ability.

(1) 6 MW: Subjects walked alone for 6 minutes wearing comfortable footwear

(2) TUG: Subjects stood up from an armchair, walked 3 meters, and then returned to the original armchair and sat down.

(3) OLS: With their hands on the hips, subjects stood with their feet together and their eyes open (EO). The subjects then closed their eyes and attempted to maintain balance while standing on their right or left leg. The subjects stood until they lost their balance, or reached a maximum testing period of 60 s. Subjects were barefoot for this test.

COP Signals

COP signals were collected and divided into the AP and ML direction signals using $Y_0(t)$ and $X_0(t)$, $t = 1,...N$. In this study, signals were collected for a time (T) of 60 s and the sample length (N) was 6000. These signals

were defined as COP-based raw data relative to the mean of the original source, $\overline{Y}(t)=\frac{1}{N}\sum Y_0(t)$ and $\overline{X}(t)=\frac{1}{N}\sum X_0(t)$, then imported into $Y(t)=Y_0(t)-\overline{Y}(t)$ and $X(t)=X_0(t)-\overline{X}(t)$. A previous study suggested that the effects of visual input should also be considered [35]; therefore, we analyzed COP signals with eyes open and closed.

The COP displacement signal is often collected, normalized, and used to examine postural stability from a force plate. The traditional COP indicators used in this study included time-domain distance parameters [12] such as the average distance, mean resultant distance (MDIST), root mean square distance (RDIST), and average speed (MVELO). The 95% confidence circle area (AREA-CC) and 95% confidence ellipse area (AREA-CE) provided estimates of the postural sway area used to estimate either the AP or ML displacement of the COP data. The included parameters are calculated as follows:

(1) Mean resultant distance (MDIST) $=\frac{1}{N}\sum_{t=1}^{N}RD(t)$ where $RD=\sqrt{X(t)^2+Y(t)^2}$.

(2) Mean resultant distance anterior– posterior (MDIST_AP) $=\frac{1}{N}\sum_{t=1}^{N}\left|Y(t)\right|$.

(3) Mean resultant distance medial– lateral (MDIST_ML) $=\frac{1}{N}\sum_{t=1}^{N}\left|X(t)\right|$.

(4) Root mean square distance (RDIST) $=\sqrt{\frac{1}{N}\sum_{t=1}^{N}RD^2(t)}$.

(5) Root mean square distance anterior– posterior (RDIST_AP) $=\sqrt{\frac{1}{N}\sum_{t=1}^{N}Y^2(t)}$.

(6) Root mean square distance medial– lateral (RDIST_ML) $=\sqrt{\frac{1}{N}\sum_{t=1}^{N}X^2(t)}$.

(7) Mean displacement velocity (MVELO) $=\frac{\text{TOTEX}}{T}$, Total excursion (TOTEX) $=\sum_{t=1}^{N-1}\sqrt{[Y(t+1)-Y(t)]^2+[X(t+1)-X(t)]^2}$.

(8) Mean displacement velocity anterior – posterior (MVELO_AP) $=\frac{\text{TOTEX\_AP}}{T}$, $\text{TOTEX\_AP}=\sum_{t=1}^{N-1}|Y(t+1)-Y(t)|$.

(9) Mean displacement velocity medial– lateral (MVELO_ML) $= \frac{\text{TOTEX\_ML}}{T}$, $\text{TOTEX\_ML} = \sum_{t=1}^{N-1} | X(t+1) - X(t) |$.

(10) 95% confidence circle area (AREA-CC) $= \pi(\text{MDIST} + z_{0.05} \times S_{RD})^2$ where $S_{RD} = \sqrt{\text{RDIST}^2 - \text{MDIST}^2}$.

(11) 95% confidence ellipse area (AREA-CE) $= \pi ab = 2\pi F_{0.05,2,t-2}\sqrt{s_X^2 s_Y^2 - s_{XY}^2}$,

where $a = \sqrt{F_{0.05,2,t-2}(s_X^2 + s_Y^2 + D)}$, $b = \sqrt{F_{0.05,2,t-2}(s_X^2 + s_Y^2 - D)}$, $D = \sqrt{(s_X^2 + s_Y^2) - 4(s_X^2 s_Y^2 - s_{XY}^2)}$, and $s_{XY} = \frac{1}{N}\sum_{t=1}^{N} Y(t)X(t)$.

Empirical Mode Decomposition

Huang et al.[14] proposed the Empirical Mode Decomposition (EMD) method. Time-frequency EMD is a new data-driven signal decomposition method for nonlinear and non-stationary signals. After an original time domain signal is decomposed through EMD into n intrinsic mode functions (IMFs), the decomposed signals are represented in the form of frequencies as $IMF_i, i = 1,...n.$ (Figure 2). IMFs, in combination with a mean trend function may reflect the overall trend of the original signals. In this study set $n = 8$, the signal in the ML direction was determined as $X(t) = \sum_{i=1}^{n} IMFx_i(t) + r_x(t)$; $X(t)$. Signals that contained residuals $r(t)$ were then deleted. The signal in the AP direction was then determined using the same decomposition method: $Y(t) = \sum_{i=1}^{n} IMFy_i(t) + r_y(t)$.

Multiscale Entropy

The COP time-series data were divided into the AP and ML directions, decomposed into IMFs using EMD, then analyzed by calculating the CI using the MSE method. The MSE method is based on the sample entropy (SampEn) calculation: SampEn (m, r, N), where m is the length of sequences to be compared, r is the tolerance for accepting matches, and N is the length of the time-series. The tolerance for accepting matches is r × SD; the standard deviation of the data set [36]. Given a one-dimensional discrete time-series, the method then constructs multiple coarse-grained time-series data using the scale factor. The CI of the MSE method was defined as the area under the curve for each coarse-grained time-series (see Figure 3). In this study, scale = 20, m = 2, r = 0.15 and N= 6000 for the complexity analysis of the COP signals; $\text{CI} = \sum_{i=1}^{scale} \text{SampEn}(i)$.

Some COP signal studies have included evaluations that use different combinations of the IMFs and parameters of the MSE. Costa et al. found significant differences in complexity between older and younger adults by adding together the five highest-frequency IMFs [13]. Wei et al. and Jiang et al. also revealed a difference in combinations of IMFs for older and younger adults under different conditions [15,16]. Different parameters of MSE and different IMFs of COP signals can generate different postural stability assessment results using the MSE method. MSE methods can also be used to analyze electroencephalogram (EEG) signals to detect risk or to classify signals [37, 38].

Signal Energy

SE values were calculated according to the length of the signal amplitude in a discrete space [20]. In this study, the SE was calculated using the IMF signal, represented as $E_i = \|IMF_i(t)\|^2$, and total signal energy is represented as $tE = \sum_i E_i$, $i = 1,...n$. The IMF signals were from the decomposed COP signal, which was transferred with a mean of zero. Therefore, the sum of the squared amplitudes of the signal is equal to the variance of the normalized magnification. Namely, the SE method compares variability to explore the subject' s postural sway through observed variances in detrended IMF signals.

-Experiment Procedure and Data analysis

All subjects performed two tasks: gait function test data collection and COP signal collection. COP signal collection required subjects to stand on a force plate in a static position; they were instructed to stand quietly with arms at their sides in a naturally straight posture and look straight ahead with EO or EC randomly five times for 60s. The subjects rested for approximately 30s between each trial to avoid fatigue and prevent interfering factors affecting the results.

Figure 4 shows the experimental and analytical procedure used in this study. The quantitative postural stability analysis of the gait function test (task 1) used with the raw data. The collected COP time-series signals (task 2) were divided into the AP and ML directions then applied to the different quantitative postural stability assessments. Furthermore, the signals were detrended using EMD and then used in MSE and SE analyses. These measurement algorithms were implemented using Matlab 2009. Finally, postural stability

was verified by comparing the results of the commonly-used gait function test, traditional COP indicators, and MSE and SE values from subjects with normal and flat feet.

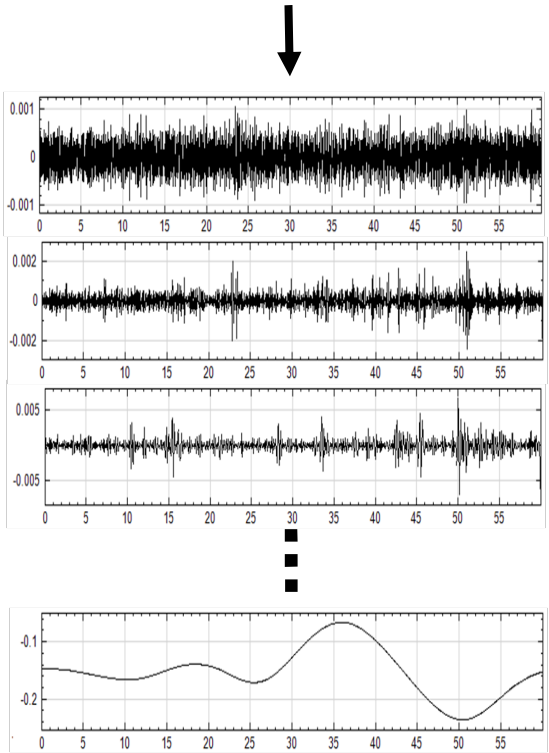

Figure 2. The time series for the COP signal was divided into X(ML) and Y(AP) direction signals, normalized, then detrended into intrinsic mode functions (IMFs) using the empirical mode decomposition (EMD) method. See (A) the COP displacement trajectory signal of X in ML direction when a subject (ID = 5) was standing on a force plate, (B) normalized trajectory signals with the mean transfer process, and (C) the EMD method decomposed into frequency IMFs.

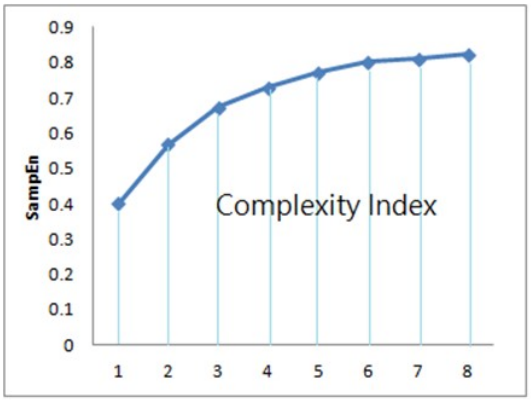

Figure 3. The CI of the MSE was defined as the area under the curve for each coarse-grained time series.

-Statistical Analysis

Differences in postural stability between normal and flat feet were analyzed using descriptive statistics with mean and standard deviation (mean ± SD). The SE results did not exhibit normal variability because the SD between the two groups was high; therefore, data were analyzed using the nonparametric Mann-Whitney test. In this study, two-sample *t*-tests were performed for the gait function test, traditional COP indicators, and the CI of the MSE method. An alpha value of 0.05 was used for all statistical tests, which were conducted using Minitab software.

RESULTS

-Gait Function Test

Table 1 shows the gait function test analysis of subjects in the flat foot and normal foot groups. The results of the 6MW and TUG showed no significant difference; only the OLS test results with EC differed significantly. In the 6MW, the average distances covered by subjects in the flat foot and normal foot groups were 511.55 and 534.29 m, respectively. The average TUG of the flat foot and normal foot groups were 8.56 and 7.99 s, respectively. The average OLS test result for the right foot of the flat foot and normal foot groups were 11.12 and 12.76 s, respectively. For the left foot, the average time of the flat foot and normal foot groups were 7.19 and 14.42 s, respectively. The average OLS test time of subjects with normal feet was significantly longer than those with flat feet when standing on the left foot ($p < 0.05$).

Although subjects in the normal foot group demonstrated higher performance in the gait function test compared with the flat foot group, only the left foot in the OLS test differed significantly between the groups.

-Traditional COP-based Parameters

Table 2 shows the p-values from traditionally-used COP-based parameters, such as MDIST, RDIST, MVELO, AREA-CC, and AREA-SW, which can be used to detect static postural stability [12], for the normal and flat foot groups with EO and EC.

The differences in the EC results between the two groups were not significant. For EO, the results revealed significant differences for the MDIST, MDIST_ML, RDIST, and RDIST_ML parameters ($p < 0.05$);

the greatest difference occurred in the ML direction. The mean average MDIST_ML and the mean average RDIST_ML values in the normal and flat foot groups differed significantly ($p < 0.05$).

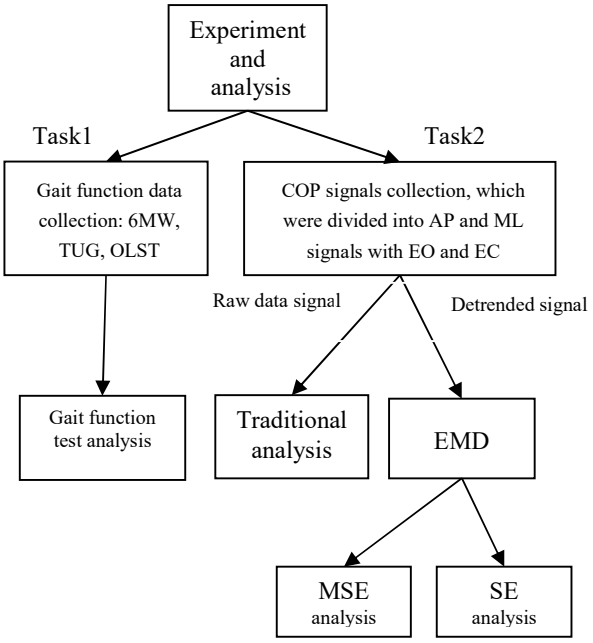

Figure 4. Schematic of postural stability assessment from the gait function test and COP displacement trajectory of subjects standing on a force plate.

Items	Flat foot (n = 17)	Normal foot (n = 37)	p-value
6MW (m)	511.55 ± 107.89	534.29 ± 82.92	0.402
TUG (s)	8.56 ± 0.51	7.99 ± 1.88	0.271
OLS (s):			
Right side	11.12 ± 8.58	12.76 ± 10.08	0.565
Left side	7.19 ± 7.15	14.42 ± 10.76	0.015*

*$p<0.05$

Table 1. Descriptive statistics (mean ± SD) for the measures of the gait function test results with groups of flat and normal feet. 6MW: 6-min walk; TUG: timed up and go; OLS: one-leg standing

-The MSE Method

Table 3 shows the p-values for the normal and flat foot groups of different IMFs with EO and EC in the AP

and ML directions. The differences in these parameters between the two groups with EO were not significant; however, a significant difference was found when the signals were detrended into IMF6 and IMF7 for EC in the AP direction ($p < 0.05$).

The results revealed a significant difference between the flat foot and normal foot groups in low-frequency signals (IMF6 to IMF7 below approximately 0.1 to 0.3 Hz) in the AP direction with EC. Furthermore, according to a previous study, IMFs can be recombined into different signals to determine optimal combinations of IMFs for postural stability analysis [15-17].

-The SE Method

In general, subjects with flat feet used more energy than those with normal feet. No significant difference in energy consumption was observed between the groups in the AP direction (Table 4). Regarding the SE results for each IMF in the ML direction: the energy (E_ML_IMFi) in both groups with EO differed significantly for the IMF1, IMF3, IMF5, IMF6, IMF7, and IMF8 signals ($p < 0.05$); moreover, there was a significant difference in signal IMF7 with EC.

Furthermore, to maintain postural stability, the total signal energy (tE) used by subjects in the flat foot group (tE_ML) was 30366.88 with EO and 50494.85 with EC. The tE used by subjects in the normal foot group was 11740.21 with EO and 19551.35 with EC. The value of subjects with flat feet was approximately twice that of subjects with normal feet, and a significant difference was observed between the groups; the differences in tE_ML were significant for EO ($p < 0.05$) and EC ($p < 0.05$).

DISCUSSION

The purpose of this study was to characterize the effectiveness of the SE method in assessing posture stability compared to other commonly used parameters. The SE method has not been previously used as a balance assessment in this field. The SE signal was squared due to the enlargement process, which enabled the exploration of variance. The results revealed that the SE-derived sensed energy values of subjects in the flat foot and normal foot groups differed significantly.

The results of this study were compared with those of other commonly used quantitative postural stability

COP-based Parameters	EO p- value	EC p-value
MDIST	0.030*	0.102
MDIST_AP	0.077	0.106
MDIST_ML	0.021*	0.192
RDIST	0.034*	0.1
RDIST_AP	0.087	0.094
RDIST_ML	0.017*	0.171
MVELO	0.572	0.54
MVELO_AP	0.339	0.512
MVELO_ML	0.985	0.723
AREA_CC	0.075	0.141
AREA_SW	0.081	0.135

*$P<0.05$

Table 2. Traditional COP-based parameters used to determine postural stability with eyes open (EO) and eyes closed (EC)

COP-based Parameters	EO p- value	EC p-value
CI_AP_IMF1	0.105	0.098
CI_AP_IMF2	0.16	0.954
CI_AP_IMF3	0.357	0.829
CI_AP_IMF4	0.655	0.368
CI_AP_IMF5	0.711	0.544
CI_AP_IMF6	0.245	0.036*
CI_AP_IMF7	0.549	0.036*
CI_AP_IMF8	0.458	0.899
CI_AP_ALL IMF	0.129	0.567
CI_ML_IMF1	0.695	0.201
CI_ML_IMF2	0.332	0.081
CI_ML_IMF3	0.599	0.507
CI_ML_IMF4	0.65	0.346
CI_ML_IMF5	0.705	0.719
CI_ML_IMF6	0.188	0.095
CI_ML_IMF7	0.129	0.312
CI_ML_IMF8	0.244	0.293
CI_ML_ALL IMF	0.053	0.305

*$P<0.05$

Table 3. MSE analysis in AP and ML directions with eyes open (EO) and eyes closed (EC)

COP-based Parameters	EO p- value	EC p-value
E_AP_IMF1	0.434	0.681
E_AP_IMF2	0.09	0.334
E_AP_IMF3	0.097	0.451
E_AP_IMF4	0.212	0.493
E_AP_IMF5	0.071	0.198
E_AP_IMF6	0.109	0.331
E_AP_IMF7	0.315	0.239
E_AP_IMF8	0.723	0.574
tE_AP	0.174	0.174
E_ML_IMF1	0.041*	0.071
E_ML_IMF2	0.412	0.233
E_ML_IMF3	0.044*	0.131
E_ML_IMF4	0.118	0.576
E_ML_IMF5	0.042*	0.055
E_ML_IMF6	0.008*	0.077
E_ML_IMF7	0.046*	0.009*
E_ML_IMF8	0.001*	0.058
tE_ML	0.004*	0.044*

*p<0.05

Table 4. SE analysis in the AP and ML directions with eyes open (EO) and eyes closed (EO) for each energy value(E) of IMFi and the total energy (tE).

parameters. These findings were supported by the statistically significant differences in the SE method results compared with those of the other parameters.

1. Regarding the gait function test items with raw data, in the OLS test when standing on the left foot with EC, the two groups differed significantly (Table 1).
2. The collected COP signals were verified using traditionally used measurements; MDIST and RDIST differed significantly in the ML direction with EO (Table 2).
3. CI values determined using the human non-stationary MSE method with the same COP signals differed significantly at low-frequencies (IMF6 and IMF7) in the AP direction with EC (Table 3) without recombining IMFs.
4. The energy (E) of each IMFi component was observed difference in the high and low frequency signals, and the energy levels of these components differed significantly in the ML direction with EO (Table 4).

Furthermore, the tE of subjects with normal feet was differed significantly from subjects with flatfeet when attempting to maintain postural stability in the ML direction with EO and EC.

The SE method results obtained in the ML direction with EC and EO differed significantly between the flat foot and normal foot groups. The frequently used COP-based parameters, MDIST and RDIST, differed significantly between the two groups in the ML direction with EO, which is similar to the results of a previous study that compared the parameters between young and elderly subjects [12].

The CI values obtained using the MSE method differed significantly between the flat foot and the normal foot groups in the AP direction with EC. The CI represents the ability to adapt to environmental change; a high CI value means high adaptability. Therefore, according to the results, the postural stability of subjects with normal feet was higher than that of subjects with flat feet. When using the MSE method, exploration of appropriate parameters and recombined IMFs to determine the CI value of the complexity calculating algorithm may improve the ability to differentiate between different groups [16, 36, 39]. Previous studies applying the MSE method have revealed that significant differences in postural stability were observed mainly in high frequency signals (the first five IMFs, for frequencies ranging from approximately 1 to 30 Hz) [13,40]. Furthermore, low frequency signals may be more sensitive for assessing differences between groups of young subjects [10, 17]. When different frequency signal combinations were created and the appropriate adapted parameters were chosen, the differences were more pronounced when eyes were closed compared with open [39]. This indicates that the SE method is straightforward because it does not entail exploring all parameters and IMFs.

The findings of this study are similar to with previously published studies about energy cost in flatfeet. Subjects with flat feet required more energy than those with normal feet. Previous studies have indicated that flat feet might incur a higher energy cost than feet with an arch support during walking [4], and the energy consumption of people with flat feet is higher than that of people with normal feet during walking [19]. Metabolic energy consumption during walking and quiet standing is related to the muscle force and muscle work required to support the body and the effort required for balance control [4, 18]. Hence, people with flat feet consume a higher amount of energy during compensatory walking movements or quiet standing to

control postural stability. In this study, the tE consumed by subjects in the flat foot group was significantly higher than that consumed by subjects in the normal foot group. This may be because the arched structure of flat feet leads to uneven plantar pressure forces, which results in a higher degree of shaking compared with normal feet, and thus requires more swaying to maintain postural stability.

In this study, the COP displacement signals of subjects standing upright on a force plate were collected and divided into AP and ML directions. Nonlinear EMD was then applied to reconstruct non-stationary time-series IMF signals, which were imported into the formulas of the SE method to observe and analyze the energy of subjects with flat foot or normal foot structures. The SE method reflected differences in the postural stability of the subjects, especially in the ML direction when standing upright with eyes both open and closed. Although the contributions of postural stability in the AP and ML directions were examined, a recent study has shown that the effects of aging on balance may be accentuated in the ML direction [41]. This earlier study suggested that amplitude in the ML direction maybe related to the increased risk of falls associated with increased age. Therefore, further study of subjects from different age groups is required to explore the application of the SE method in postural stability in the ML direction.

The SE method proposed in the present study may be used in future clinical practice to determine responsive changes in the postural stability of people with flat feet after treatment. According to the results, SE should be considered to be a useful outcome measure of balance control. The SE method can be applied to a low cost, portable device to provide healthcare services during daily life activities and prevent the occurrence of falls in older people with flat feet.

CONCLUSIONS

The findings of this study indicated that amplification and variance exploration are the main characteristics of signal energy, which can enable differentiation between flat and normal feet. The SE method proposed in this study could be used for sensed balance measurements in future clinical practice to determine changes in the postural stability of people with flat feet after treatment to enhance postural stability, such as corrective aids, sports training and other clinical interventions. Helping clinicians find subtle changes in postural stability,

while treating the symptoms of flat feet is challenging. In the future, the proposed SE method may help clinicians to identify the effectiveness of strategies and apply the method to other biological time-series data.

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