Comparison of Different Structural Foot Types for Measures of Standing Postural Control

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Study Design: Matched group comparison of 3 subject groups with 3 different foot structures for force plate and clinical measures of postural control.

Objectives: To determine if subjects with different weight-bearing foot structure would demonstrate differences in static standing postural control, and to determine the reliability of study procedures.

Background: Weight-bearing foot structure may influence postural control either because of a decreased base of support (supinated foot structure) or because of passive instability of the joints of the foot (pronated foot structure).

Methods and Measures: Young adults were categorized based on weight-bearing foot structure into neutral, pronated, or supinated groups (15 subjects per group). Postural control in single-limb stance with eyes closed was assessed using force plate measures and by measuring duration of single-limb stance on a firm floor and on a balance pad. Force plate measures were normalized center-of-pressure average speed; and standard deviation and maximum displacement in the anterior-posterior and medial-lateral directions.

Results: Individuals in the supinated group had significantly greater center-of-pressure average speed, greater maximum displacement in the anterior-posterior direction, and greater SD and maximum displacement in the medial-lateral direction than individuals in the neutral group. The individuals in the pronated group had significantly greater SD and maximum displacement in the anterior-posterior direction, used more trials to complete force plate testing, and had shorter single-limb stance duration than those in the neutral group.

Conclusion: Individuals with pronated feet or supinated feet have poorer postural control than individuals with neutral feet, but perhaps through different mechanisms. *J Orthop Sports Phys Ther 2006;36(12):942-953.* doi:10.2519/jospt.2006.2336

Key Words: *balance, feet, pronation, supination*

Static postural control, or steadiness, can refer to the ability to hold the body as motionless as possible for a given condition and position.[10] Specifically, static postural control can be defined as the ability to stabilize or minimize the movement of the center of gravity within the base of support when equilibrium status is achieved for a given condition and weight-bearing position.[12,24,39] Because of the difficulty in locating the center of mass of the body, the most common laboratory method for assessing static standing postural control involves using a force plate to record an individual's center-of-pressure trajectory for a standing test position during a period of time.[15,28] Movements of the center of pressure reflect the body's efforts to adjust and position the center of mass over the base of support.[12,27,40] An individual with a greater center-of-pressure total path length during a particular task, therefore, may possess poorer postural control than an individual with a lesser center-of-pressure total path length.[24] If the testing periods are not identical across subjects and testing trials, standing postural control can be standardized by dividing the total travel distance of the center

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of pressure by the testing period to represent the average speed of the center-of-pressure travel path.[27]

Previous researchers have reported that subjects who had poorer preseason measures of postural control were more likely to incur ankle injuries during a season of soccer play[36] or were more likely to incur ankle sprain injury during a basketball season.[21] Several authors have emphasized the importance of balance training in rehabilitation or sports training programs to restore normal postural control function and to prevent recurrent or initial injury.[11,14,35,37,38] Impaired postural control has also been related to an increased risk of fall injury in the elderly population.[19,33,34] A better understanding of the variables that influence postural control may be helpful in the prevention of injuries.[3]

An area of postural control that has not been well investigated is the influence of foot structure on postural control. The supinated (high-arched) foot may be associated with excessive subtalar joint supination, theoretically reducing the plantar contact area with the standing surface.[8] The reduction in contact area for the supinated foot may reduce sensory input from plantar sensory end organs,[13] thereby reducing sensory input that is important for controlling balance.[20] Hertel et al[13] have presented data indicating that subjects with a supinated foot structure demonstrated significantly greater center-of-pressure excursion velocity compared with subjects with a neutral foot structure for a 10-second, barefoot, single-leg-stance condition with eyes open.

The supinated foot structure may also be associated with generalized hypomobility of joints within the foot,[5] making adjustments to center-of-mass perturbations more challenging. Theoretically, the body may rely more on the movements of the proximal joints of the lower extremity (eg, hip and knee joints) for controlling unilateral upright stance on a hypomobile foot. Movements of the proximal joints may elevate portions of the hypomobile foot from the support surface more easily, as the foot joints may be too inflexible to adapt to the standing surface. Elevation of portions of the foot during postural control adjustments may then result in a narrower base of support, and standing with a smaller base of support tends to increase the challenge to postural control.[2,18,22]

The pronated (flat-arched) foot may be associated with excessive subtalar joint pronation. Abnormal compensatory foot pronation may cause passive instability and hypermobility of the joints of the foot.[8] The pronated foot, therefore, may be unstable during weight bearing and might impair postural control.[3] Cobb et al[3] reported that subjects with bilateral forefoot varus greater than or equal to 7° had a significantly poorer anterior-posterior (AP) postural stability (standard deviation of the AP ground reaction force) than the group with forefoot varus less than 7° for a 5-second, barefoot, single-leg-stance balance task with eyes closed. Forefoot varus deformity assessed during non-weight bearing, however, may not always be associated with compensatory foot pronation during weight bearing. Cote et al[4] have also reported significantly poorer stability (mean sway deviation around the center of balance) for a group of subjects with pronated feet compared to a group of subjects with supinated feet for static postural control testing, using the Chattecx Balance System during 15-second single-limb stance with eyes open and eyes closed. Foot classification was based on the navicular drop test for this study.[4]

Comparing the results of the previously reviewed studies[3,4,13] is difficult because of the differences in the number of foot structure groups, foot classification criteria, and the variables used to assess postural control. All of these studies have used foot classification methods that involve the determination of subtalar-neutral position, an assessment that has poor intratester and intertester reliability.[7,25,30,32] Another very important issue for some of these studies[3,13] involves the use of a non–weight-bearing assessment of foot structure. The structure of the foot observed in non-weight bearing may not guarantee specific structural appearance of the foot under weight-bearing conditions.[30] The purpose of this study, therefore, was to determine if subjects with different weight-bearing foot structure would demonstrate differences in static postural control during single-limb stance. A second purpose of our work was to determine the reliability of some of the study procedures.

METHODS

Subjects

Three subject groups participated in the study and consisted of individuals with pronated, neutral, or supinated feet, based on a weight-bearing foot classification scheme. Each group was comprised of 8 males and 7 females between 18 and 31 years of age and no history of lower extremity injury during the 6 months prior to participation in the study. Only individuals with symmetric left and right foot structure were recruited. Subjects also reported that they had no diagnosis of neural or vestibular disease or lower extremity arthritis.

At the time of participation in the study, subjects were asked to complete a questionnaire regarding injury history, pain, and use of drug and dietary substances. Individuals were excluded from the study if any pain was present in the lower extremities and if any substance that might affect postural control (eg, alcohol, sedatives, cold remedies, stimulants, etc) had been taken within the last 24 hours. Subjects' physical status in terms of exercise time and type was also recorded on the questionnaire. To strengthen the internal validity of the study, subjects who had engaged in exercise or training that might require good postural control ability (eg, ballet, gymnastics, Tai Chi, etc) during 1 year prior to participation in the study were excluded from the study. Subjects who had engaged in these kinds of exercise or training for a total period of more than 1 year during the past 10 years were also excluded from the study. The study protocol was approved by The Biomedical Internal Review Board at The University of North Carolina at Chapel Hill, and all subjects read and signed a statement of informed consent.

Classification of Weight-Bearing Foot Structure

Weight-bearing foot structure was classified as pronated, neutral, or supinated, using the weight-bearing foot assessment adapted from the original foot classification scheme proposed by Jonson and Gross.[17] Jonson and Gross[17] reported acceptable intratester (intraclass correlation coefficient [$ICC_{2,1}$] = 0.88-0.90) and intertester ($ICC_{2,1}$ = 0.81 to 0.86) reliability for the assessments of medial longitudinal arch angle and rearfoot-to-leg angle to classify foot type.[17] The medial longitudinal arch angle (Figure 1) is the obtuse angle between the line connecting the medial malleolus and the navicular tuberosity and the line connecting the navicular tuberosity and the most medial aspect of the first metatarsal head. The rearfoot-to-leg angle (Figure 2) is the acute angle formed by the longitudinal bisecting line of the calcaneus and the longitudinal bisecting line of the distal one third of the leg.

Medial longitudinal arch angle and rearfoot-to-leg angle were measured for both feet of each subject. Each subject was positioned in relaxed bilateral standing, with the distance between the 2 ankle joint centers equal to the distance between the right and left anterior superior iliac spines (Figure 3). To draw the 2 bisecting lines for the rearfoot-to-leg angle measure, a standardized scheme proposed by Genova and Gross[9] was adopted to locate the midpoint marks on the calcaneus and the distal one third of the leg (Figure 2). A foot was classified as pronated if the rearfoot eversion angle was greater than 9° and the medial longitudinal arch angle was less than 134°. A foot was classified as supinated if the rearfoot eversion angle was less than 3° and the medial longitudinal arch angle was greater than 150°. A foot was classified as neutral if the rearfoot eversion angle was between 3° and 9° and the medial longitudinal arch angle was between 134° and 150°.

The reliability of the procedures for measuring the medial longitudinal arch angle and rearfoot-to-leg angle was investigated for 14 feet in a previous pilot study. Subjects who participated in the pilot study did not contribute data for the current study. Reliability was evaluated using the $ICC_{3,1}$ and mean absolute difference computations. Intertester reliability was based on the same 2 investigators, each making a single measurement of both variables on each of the 14 feet. Intertester reliability for the medial longitudinal arch angle was 0.89, with a mean absolute difference of 3.9°. Intertester reliability for the rearfoot-to-leg angle was 0.81, with a mean absolute difference of 2.6°. Intratester reliability was based on the same investigator making 2 measurements of each variable on each of the 14 feet. Intratester reliability for the medial longitudinal arch angle was 0.94, with a mean absolute difference of 3.0°. Intratester reliability for the rearfoot-to-leg angle was 0.86, with a mean absolute difference of 1.3°.

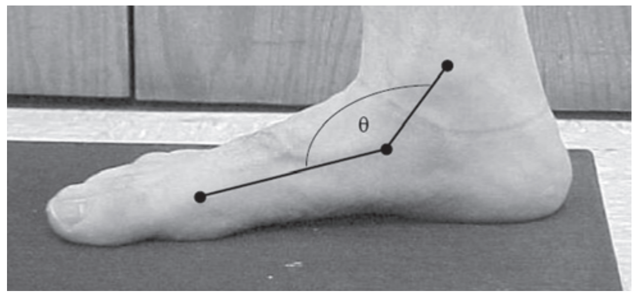

FIGURE 1. Measurement of the medial longitudinal arch angle, the obtuse angle (θ) between the line connecting the medial malleolus and navicular tuberosity, and the line connecting the navicular tuberosity and the most medial aspect of the first metatarsal head.

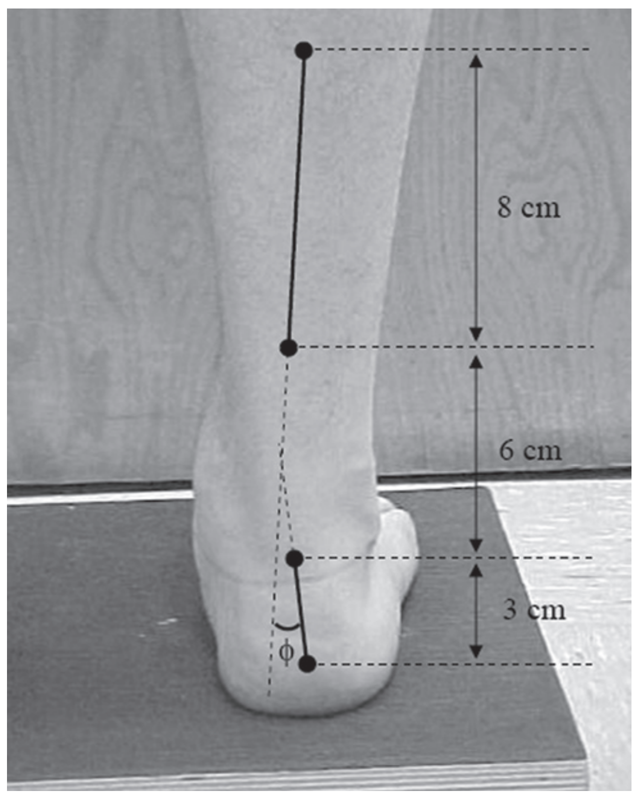

FIGURE 2. Measurement of the rearfoot-to-leg angle (ϕ) and the standardized scheme for locating midpoint marks on the calcaneus and on the distal one third of the leg. The rearfoot-to-leg angle is the acute angle (ϕ) between the bisecting line of the calcaneus and the bisecting line of the distal one third of the leg.

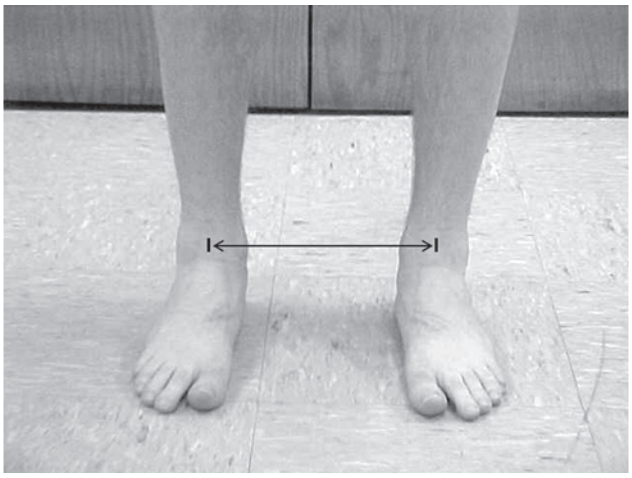

FIGURE 3. Standardized scheme for positioning the distance between the subject's feet. Each subject's feet were separated so that the distance between the 2 ankle joint centers was equal to the distance between the right and left anterior superior iliac spines.

We hypothesized that the increase in passive instability of joints in the pronated foot might impair standing postural control. Subjects with rigidly pronated feet, therefore, were excluded from the pronated-foot group. A rigidly pronated foot generally does not change its appearance when an individual increases the amount of weight bearing. We recruited subjects for the pronated-foot group who demonstrated foot pronation movement as their weight-bearing status changed from minimum weight bearing to 50% weight bearing. The medial longitudinal arch angle of the subjects in the pronated-foot group was measured with subjects positioned in relaxed bilateral stance and with subjects seated with hip and knee joints positioned in 90° of flexion and the trunk in an erect position. The difference between the medial longitudinal arch angles for the 2 test positions was calculated to determine whether sufficient foot pronation movement was present as the subject changed his or her weight-bearing status. For an individual to be included in the pronated-foot group, the difference between the medial longitudinal arch angles for the 2 testing positions had to be greater than or equal to 8°. The value of 8° represents 1 standard deviation for the sample distribution reported by Jonson and Gross[17] for the medial longitudinal arch angle measured with subjects positioned in bilateral stance.

Force Plate Measures

Two Bertec 4060A force plates (Bertec, Worthington, OH) were used to collect 3-dimensional ground reaction force data at 30 Hz. The trajectory of the center of pressure was calculated by a custom computer software program. Static postural control in single-limb stance was assessed using the force plate measures of average speed of the center of pressure (total path length per second) and the standard deviation and maximum displacement of the center of pressure in the AP and medial-lateral (ML) directions. Average speed of the center of pressure was used as a study variable to standardize the center-of-pressure total path length, recognizing that all subjects might not stand for the same period of time during the test trials. The origin of the center-of-pressure path was the initial point of the center of pressure during each trial. The software program then computed total path length of the center of pressure for the trial by summing the displacement of the center of pressure between successive sample points for the entire trial. The average speed of the center of pressure was then computed by dividing the total path length by the duration of the trial.

Because 1 force plate measure alone may not be able to represent adequately the various aspects of postural control, we also measured the standard deviation and the maximum displacement of the center of pressure to reflect the variability and the range of displacement of the center of pressure, respectively.[27] The standard deviation of the center of pressure about the mean value and maximum displacement of the center of pressure in the AP and ML directions were derived from the center-of-pressure trajectory data. Maximum displacement of the center of pressure in the AP direction was computed as the difference between the maximum anterior position and the maximum posterior position of the center of pressure. Maximum displacement of the center of pressure in the ML direction was computed as the difference between the maximum medial position and the maximum lateral position of the center of pressure. The standard deviation and maximum displacement variables were used to gain additional information about subjects' static standing postural control, because the 2 subjects could have identical average speed values with very different displacement profiles (eg, higher frequency smaller oscillations, compared with lower frequency larger oscillations, of the center of pressure).

Each subject performed 10-second trials of barefoot single-limb stance with eyes closed for the right and left lower extremities on the force plates. Subjects were positioned with eyes closed and with their hands on their iliac crests. Pilot testing indicated that testing with eyes open was not sufficiently challenging to discriminate among individuals. The test limb was in full knee extension, with toes toward the anterior direction of the force plates, and the foot of the nontest limb was positioned above the test surface without touching the test limb.[10] Each subject was asked to perform the required 10-second single-limb stance as motionlessly as possible. Subjects were given 3 practice trials and 3 test trials for each limb, with 1-minute rest periods between consecutive trials. The test order between limbs was counterbalanced for

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subjects in each group. The time that the subject could maintain the required single-limb stance on the force plate was measured using a stopwatch. The stopwatch was started at the same time the force plate data collection was triggered. The stopwatch was only used, however, to determine when the subject had completed the 10-second testing trial and as an approximate indicator that the subject had met the minimum stance time of 5 seconds for an acceptable trial. The trial was stopped and the stance time (from the stopwatch) for the trial was recorded if any of the following faults occurred: (1) the foot of the nontest limb touched the testing surface or the test limb, (2) the entire foot of the test limb momentarily lifted off the testing surface, (3) the subject's eyes opened, or (4) either hand left its position on the iliac crests.

An acceptable trial required that the subject maintain the required position for a minimum of 5 consecutive seconds. If a subject was unable to complete 3 full 10-second trials during the 3 test trials, the subject was given a maximum of 3 additional trials to obtain 3 full 10-second trials. If the subject was still unable to complete 3 full 10-second trials after the 3 additional trials were given, the data from the first 3 test trials that reached the minimum acceptable time of 5 seconds were selected. If the subject was not able to provide 3 trials (out of 6 possible trials) of the required single-limb stance for at least 5 consecutive seconds, the subject was excluded from the study. The center-of-pressure measures were then derived from the period that the subject was able to maintain the required testing position for the selected trials. The number of trials needed by each subject to complete 3 full 10-second trials for each limb was also recorded.

Mean values for the 3 test trials were calculated for each limb for the following force plate variables: average speed, AP and ML standard deviations, and AP and ML maximum displacements of the center of pressure. The mean value of the average speed of the center of pressure was normalized to the length of the foot. This normalization procedure was performed because total path length of the center of pressure and, therefore, average speed of the center of pressure theoretically has the potential to be greater with a larger foot because the center of pressure can travel farther in all directions prior to reaching the limits of contact with the foot. The mean AP and ML standard deviations of the center of pressure were normalized to the length and width of the foot, respectively. The same normalization procedure was applied to the mean AP and ML maximum displacements of the center of pressure. The length of the foot was measured as the distance from the most posterior aspect of the calcaneus to the most anterior aspect of the toes. The width of the foot was measured as the distance between the first and fifth metatarsal heads. For each subject, the normalized force plate measures for the right limb and left limb were summed to represent postural control of the subject for each force plate variable.

Clinical Assessment

Because force plates are not commonly available in the clinical setting, we also measured the duration of single-limb stance with eyes closed on 2 different surfaces as a clinical test of static postural control. The duration of single-limb stance has been commonly used to evaluate postural control in the clinical setting. Previous researchers have reported good test-retest reliability of the single-limb stance test for healthy subjects.[26,31] We tested single-limb stance not only on a firm floor surface, but also on a soft balance pad (Figure 4) to increase the challenge of the clinical testing, thereby possibly providing better differentiation among the 3 groups.

The Airex balance pad (Alusuisse Airex AG, Sins, Switzerland) is made of closed-cell PVC foam, with dimensions of 50 × 41 × 6 cm, and was used to create a more challenging standing surface. Subjects were asked to perform barefoot single-limb stance with eyes closed on both the floor and the balance pad for the right and left limbs. Subjects were positioned in the same single-limb stance position that was used for the force plate measurements. Subjects were asked to perform the required single-limb stance as motionless and as long as possible for a maximum of 60 seconds per trial. Subjects were given 3 practice trials and 3 test trials for each testing surface for each limb, with 1-minute rest periods between consecutive trials. The test order between limbs was counterbalanced for subjects in each group. The time that the subject could maintain the required single-limb stance was measured using a stopwatch. The trial was stopped and the stance time for the trial was recorded if any of the following faults occurred: (1) the foot of the nontest limb touched the testing surface or the test limb; (2) the foot of the test limb momentarily lifted off the testing surface; (3) the subject's eyes opened; (4) either hand left its position on the iliac crests; or (5) the subject was able to stand in the test position for 60 seconds. To compute test-retest reliability, 5 subjects were randomly selected from each of the 3 groups. These subjects performed the clinical single-limb stance test twice with a 10-minute rest period between the 2 series of tests. The order between force plate testing and clinical testing was counterbalanced for subjects in each group.

The mean duration that the subject could maintain the required single-limb stance for the 3 test trials on the floor surface and on the balance pad was calculated for each limb. The 4 mean durations (2 surfaces for 2 limbs) were summed to create a total single-limb stance score to represent the postural control of the subject across both surfaces for the clinical measurement task.

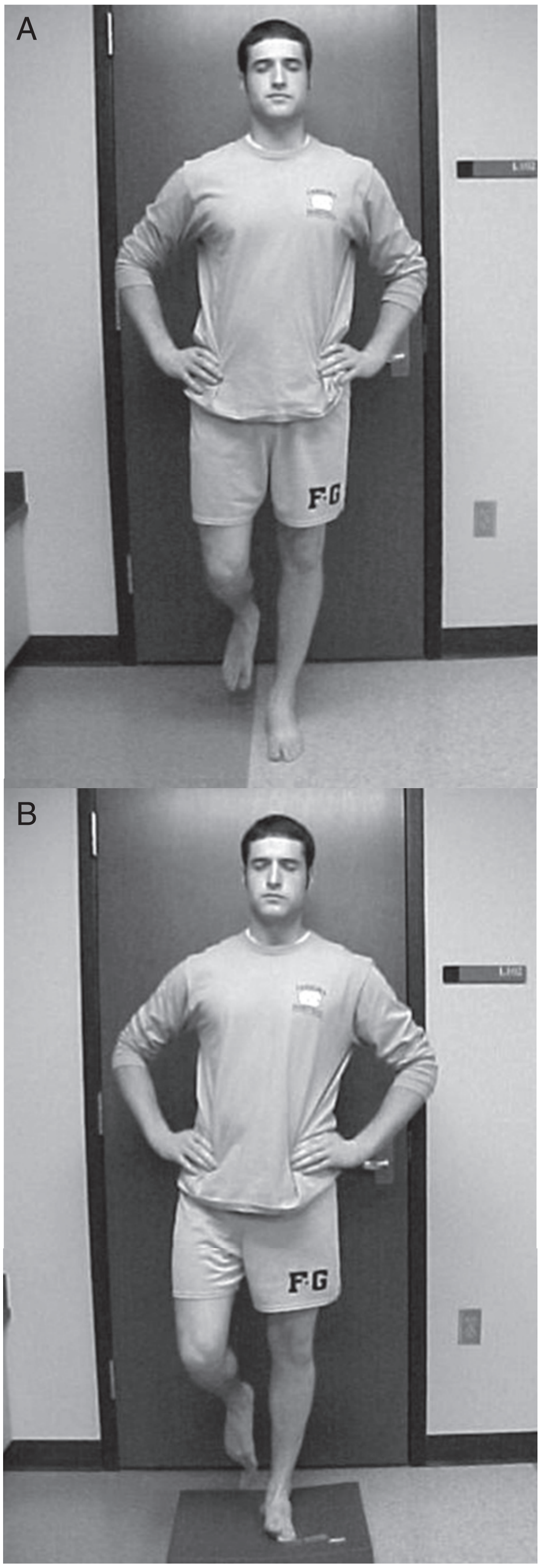

FIGURE 4. Single-limb stance on the floor surface (A) and on the Airex balance pad (B).

Data Analysis

Separate 1-way analysis of variance (ANOVA) procedures with foot type as the independent variable were used to investigate the differences in the normalized center-of-pressure average speed, standard deviation in the AP and ML directions, maximum displacement in the AP and ML directions, the total single-limb stance score among the 3 foot type groups, and all descriptive statistics for the subject groups. Post hoc analyses for multiple comparisons were performed using the Tukey honestly significantly different test. Test-retest reliability of the clinical single-limb stance test was evaluated using the intraclass correlation coefficient ($ICC_{3,3}$) and mean absolute difference computations. A Kruskal-Wallis 1-way ANOVA was used to investigate the differences in the number of trials needed to complete force plate data collection among the 3 foot type groups. The level of significance was $P<.05$ for each analysis.

RESULTS

Ninety-three healthy volunteers were screened to enroll 47 subjects who were qualified for the study based on foot classification criteria for the 3 foot structure groups. Forty-six potential subjects were excluded from the study for various reasons. Some had asymmetric right and left foot structure, based on the results of the 2 foot angle measurements, while some only qualified for the study based on 1 of the foot angle measurement criteria. Some potential subjects who met the criteria for the neutral or supinated groups were also excluded from the study because a sufficient number of subjects had already been enrolled in these 2 subject groups.

Among the 47 enrolled subjects, 2 female subjects (1 in the neutral group and 1 in the pronated group) failed to complete 3 acceptable force plate test trials for the minimum 5 consecutive seconds of single-limb stance for at least 1 limb. The data for these 2 subjects, therefore, were excluded. A total of 45 subjects' data (8 males and 7 females for each group) were then analyzed. Characteristics of the subjects for each foot structure group appear in Table 1. Subjects in the 3 groups were comparable in terms of body height, body mass, foot length, and foot width. Subjects in the neutral group were significantly older than subjects in the pronated group (26.1 years old versus 21.9 years old). Subjects in each group were significantly different from subjects in the other 2 groups with regard to medial longitudinal arch angle and rearfoot-to-leg angle.

Eight subjects in the pronated group (3 males and 5 females) and 2 male subjects in the supinated group were unable to provide 3 10-second trials of force plate data for at least 1 limb. Although not part of the original data analysis plan, we thought that this result also reflected an aspect of postural control

TABLE 1. Descriptive statistics by group. Values are mean ± 1 standard deviation. All pairwise contrasts among groups are not significant unless otherwise noted.

Variable	Neutral Group (n = 15)	Pronated Group (n = 15)	Supinated Group (n = 15)
Age (y)*	26.1 ± 3.6	21.9 ± 3.5	23.9 ± 3.2
Mass (kg)	70.5 ± 11.2	69.0 ± 11.7	67.6 ± 12.9
Height (m)	1.73 ± 0.08	1.69 ± 0.11	1.72 ± 0.09
Dominant limb			
Right	15	12	15
Left	0	3	0
Medial longitudinal arch angle (deg)			
Right†	144.0 ± 2.9	131.2 ± 1.3	157.2 ± 3.2
Left‡	143.6 ± 2.2	130.0 ± 2.1	156.3 ± 3.4
Rearfoot-to-leg angle (deg)§			
Right‖	5.7 ± 1.7	11.3 ± 1.8	–0.1 ± 1.8
Left¶	5.7 ± 1.7	12.1 ± 1.8	0.1 ± 1.9
Foot length (cm)			
Right	25.8 ± 1.6	26.0 ± 1.8	24.9 ± 1.6
Left	25.8 ± 1.5	26.0 ± 1.7	24.9 ± 1.6
Foot width (cm)			
Right	9.8 ± 0.6	9.4 ± 0.6	9.3 ± 0.7
Left	9.7 ± 0.5	9.4 ± 0.6	9.2 ± 0.6

* Neutral group significantly older than the pronated group (Tukey honestly significantly different test [HSD], 3.0 years).
† All groups are significantly different from each other (F = 373.9; *df* = 2,42; HSD, 2.3).
‡ All groups are significantly different from each other (F = 376.8; *df* = 2,42; HSD, 2.3).
§ Positive numbers are calcaneal eversion and negative numbers are calcaneal inversion.
‖ All groups are significantly different from each other (F = 161.8; *df* = 2,42; HSD, 1.6).
¶ All groups are significantly different from each other (F = 169.0; *df* = 2,42; HSD, 1.6).

during single-limb stance. The results of the Kruskal-Wallis 1-way ANOVA indicated significant differences (H = 11.64, $P<.05$) among the 3 groups for the total number of trials needed to complete force plate data collection. A Mann-Whitney U analysis with Bonferroni correction of significance level indicated that subjects in the pronated group needed significantly ($U = 34$, $P<.0167$) more trials (mean ± SD, 9.3 ± 2.1 trials) to complete force plate data collection than subjects in the neutral group (mean ± SD, 7.0 ± 1.0 trials). The number of trials needed by individuals in the supinated group (mean ± SD, 7.8 ± 1.9 trials) was not significantly different from those in the pronated group ($U = 63.5$) or the neutral group ($U = 80.0$).

The results of the 1-way ANOVA and Tukey honestly significantly different (HSD) procedures indicated significant differences among the 3 groups for all 5 of the normalized center-of-pressure variables (Table 2). Subjects in the supinated group had a significantly greater normalized center-of-pressure average speed and maximum displacement in the AP direction, and a significantly greater standard deviation and maximum displacement in the ML direction, than subjects in the neutral group. Subjects in the pronated group had a significantly greater standard deviation and maximum displacement in the AP direction than subjects in the neutral group. Subjects in the supinated group had a significantly greater maximum displacement in the ML direction than subjects in the pronated group. All other pairwise comparisons did not reveal significant differences.

Reliability results for the clinical timed single-limb stance test indicated acceptable test-retest reliability for the floor surface score and total single-limb stance score ($ICC_{3,3}$ = 0.89 and 0.91, respectively), but only moderate reliability for the balance pad score ($ICC_{3,3}$ = 0.64). The mean absolute difference between repeated test results for the total single-limb stance score was 12.45 seconds. The 1-way ANOVA for the total single-limb stance score indicated significant differences among the 3 subject groups (Table 2). Total single-limb stance scores for subjects in the pronated group (mean ± SD, 41.1 ± 20.0 seconds) were significantly less than the total single-limb stance scores for subjects in the neutral group (mean ± SD, 77.4 ± 31.1 seconds) and the supinated group (mean ± SD, 70.2 ± 26.1 seconds). The neutral group and supinated group were not significantly different for this measure.

DISCUSSION

Subjects in the neutral, pronated, and supinated groups had symmetric right and left foot structure based on the criteria of the medial longitudinal arch angle and rearfoot-to-leg angle measurements. The results of our previous pilot study also indicate acceptable intertester and intratester reliability for these 2 measures. Subjects enrolled in the present study, therefore, had distinguishable neutral, pronated, or supinated weight-bearing foot structure based on the inclusion criteria for the study. Although subjects in the neutral group were older than subjects in the pronated group, previous studies indicate no age-related differences in postural control within the age range for subjects tested in our study.[1,6]

The center-of-pressure average speed values in this study are slightly greater than values reported by Ekdahl et al,[6] who measured center-of-pressure movements of subjects using similar procedures. The mean values of the center-of-pressure average speed for the right and left limbs for the neutral group (7.7 and 8.0 cm/s, respectively) prior to our normalization procedure were slightly greater than the center-of-pressure average speed reported by Ekdahl et al.[6] Ekdahl et al[6] reported center-of-pressure average speeds for subjects in the 20- to 29-years-age group were 5.7 cm/s (right limb) and 5.9 cm/s (left limb) for males, and 5.9 cm/s (right limb) and 6.2 cm/s (left limb) for females. The center-of-pressure average speeds reported by Ekdahl et al,[6] however, were derived from force plate data only for subjects who could maintain the required single-limb stance with eyes closed for 30 seconds (appreciably longer than the 10-second period used for the present study). Average speed of the center of pressure may be less for people who have better standing postural control when the testing position is maintained for a longer period of time.

The significant group differences for the total single-limb stance score between the neutral and pronated groups and between the pronated and supinated groups may reflect valid differences between these groups. Duration of standing on the floor surface and the total single-limb stance score had good test-retest reliability ($ICC_{3,3}$ = 0.89 and 0.91, respectively). The mean absolute difference between repeated tests for the total single-limb stance score during reliability testing was 12.5 seconds, much shorter than the group differences between the neutral and pronated groups (36.3 seconds) and between the pronated and supinated groups (29.0 seconds) for this study variable.

Test-retest reliability for the duration of standing on the balance pad ($ICC_{3,3}$ = 0.64), however, was not acceptable. A learning effect might have existed for single-limb stance on the balance pad, based on qualitative observations during data collection of the clinical timed single-limb stance test. A post hoc paired *t* test was conducted to evaluate any learning effects for performing the clinical test. The results indicated that the mean duration of single-limb stance on the balance pad for the second test session (10.8 seconds) was significantly longer than the

TABLE 2. Descriptive statistics and analysis of variance results for comparison of the 3 subject groups on the study outcome variables. Values are mean ± 1 standard deviation, and represent the sum of measurements for the right and left feet of subjects. All pairwise contrasts not significantly different ($P>.05$) unless noted.

Variable	Neutral Group (n = 15)	Pronated Group (n = 15)	Supinated Group (n = 15)
Average speed, center of pressure (% of foot length)*	61.1 ± 10.7	66.6 ± 9.3	72.4 ± 14.7
Standard deviation, center of pressure, AP direction (% of foot length)†	8.6 ± 1.5	10.9 ± 1.9	10.5 ± 2.8
Maximum displacement, center of pressure, AP direction (% of foot length)‡	42.4 ± 7.3	55.3 ± 10.5	53.0 ± 16.2
Standard deviation, center of pressure, ML direction (% of foot width)§	19.8 ± 3.3	22.9 ± 3.4	24.6 ± 0.9
Maximum displacement, center of pressure, ML direction (% of foot width)‖	82.4 ± 10.4	87.3 ± 12.8	99.1 ± 14.3
Total single-limb stance time for both surfaces (s)¶	77.4 ± 31.1	41.3 ± 20.0	70.2 ± 26.1

Abbrevations: AP, anterior-posterior; ML, medial-lateral.
* Supinated group significantly greater than ($P<.05$) neutral group (F = 3.49; Tukey honestly significantly different test [HSD], 10.49).
† Pronated group significantly greater than ($P<.05$) neutral group (F = 4.74; HSD, 1.87).
‡ Pronated and supinated groups significantly greater than ($P<.05$) neutral group (F = 4.97; HSD, 10.60).
§ Supinated group significantly greater than ($P<.05$) the neutral group (F = 5.76; HSD, 3.49).
‖ Supinated group significantly greater than ($P<.05$) pronated and neutral groups (F = 6.97; HSD, 11.18).
¶ Pronated group significantly less than ($P<.05$) supinated and neutral groups (F = 8.10; HSD, 23.21).

duration for the first test session (8.0 seconds), suggesting a learning effect.

Subjects in the pronated group had poorer standing postural control than subjects in the neutral group in terms of a greater normalized center-of-pressure standard deviation and maximum displacement in the AP direction. Cobb et al[3] reported that a group with forefoot varus greater than or equal to 7° had a significantly greater AP standard deviation of the ground reaction force than a group with forefoot varus less than 7° for a 5-second single-limb stance with eyes closed. The finding that the pronated group had a significantly greater normalized center-of-pressure standard deviation and maximum displacement in the AP direction than the neutral group in the current study is in agreement with the results reported by Cobb et al,[3] although the evaluation parameters for standing postural control are different for the 2 studies.

We hypothesized that subjects with a pronated foot structure would have poorer standing postural control than subjects with neutral feet, because of reduced stability within the foot joints.[3] Subjects in the pronated-foot group, however, might also have had some advantage from the increased ML dimension of the base of support that coincides with foot pronation. The possible deleterious influence on postural control caused by passive instability in the joints of the foot, therefore, might be attenuated somewhat in the ML direction by the positive influence of an enlarged base of support. This may explain why the pronated-foot group had a significantly greater normalized center-of-pressure standard deviation and maximum displacement than the neutral group in the AP direction and not in the ML direction. The center-of-pressure average speed is a combination of the center-of-pressure movements in the AP and ML directions. A greater normalized center-of-pressure standard deviation and maximum displacement in the AP direction alone might not be sufficient for the pronated group to display a significantly greater normalized center-of-pressure average speed than the neutral group (66.6% of foot length per second versus 61.1% of foot length per second).

Subjects in the pronated group also needed significantly more trials to complete force plate data collection than the neutral group. More than half of the subjects in the pronated group (8 out of 15) could not accomplish 3 10-second trials for force plate measurements for at least 1 limb. Subjects in the pronated group also had a significantly shorter duration for the clinical timed single-limb stance test compared with the duration of the neutral group. Subjects in the pronated group, therefore, had greater difficulty accomplishing the laboratory and clinical postural control tasks of single-limb stance than subjects in the neutral group. The results further indicate that individuals with pronated feet may be at greater risk for loss of balance and falls when they are required to stand in unilateral stance for functional activities.

Subjects in the supinated group also demonstrated poorer static postural control for single-limb stance than subjects in the neutral group, based on the definition that a greater center-of-pressure movement indicates poorer postural control. Individuals in the supinated group had significantly greater normalized center-of-pressure measures, except for the standard deviation in the AP direction. Subjects in the supinated group, however, used a similar number of trials to complete force plate data collection and had similar performance for the clinical timed single-limb stance test, compared with the measures for subjects in the neutral group. Individuals with supinated feet, therefore, may not have increased risk compared with individuals who have neutral foot structure for loss of balance and falls when they are required to stand in unilateral stance for functional activities. Even though individuals with supinated feet demonstrated greater center-of-pressure displacement than individuals with neutral feet, the 2 groups of subjects had comparable ability in terms of the combined period for maintaining unilateral stance on the floor and the balance pad.

Hertel et al[13] also reported that subjects with cavus feet had a significantly larger center-of-pressure excursion area than subjects with rectus feet for a 10-second single-limb stance test with eyes open. Individuals with supinated feet in our study and in the study by Hertel et al[13] may have demonstrated greater center-of-pressure displacement than individuals with neutral feet because of the relative foot hypomobility associated with supinated foot structure.[5] Theoretically, the body may rely more on the movements of the proximal lower extremity joints for controlling single-limb stance in the presence of a hypomobile foot. Movements of the proximal joints may elevate portions of the hypomobile foot from the force plate more easily because the foot joints may not be flexible enough to adapt to the force plate. Elevation of segments of the foot during postural control adjustments may result in a narrower base of support that may then increase the challenge of postural control.[2,18,22] Unlike the pronated foot, the supinated foot may not be able to increase its ML base of support through foot pronation motion because the hypomobility of foot joints hinders the foot from pronating. Elevation of foot segments resulting from standing on a hypomobile foot, therefore, may have more critical effects on center-of-pressure movements in the ML direction than in the AP direction.

The results of the present study were somewhat contradictory to the results reported by Cote et al.[4] Cote et al[4] reported that neither the pronated group nor the supinated group in their study was significantly different from the neutral group for static postural control tests in single-limb stance. The methods used to classify foot structure and the measures used to assess static postural control were appreciably different in the 2 studies. An important contribution of the study by Cote et al[4] is their inclusion of a dynamic balance task, because standing postural control should be studied under both static and dynamic conditions to simulate functional requirements.[28] Because dynamic postural control may provide useful information for understanding underlying motor control mechanisms,[16] we recommend that future studies use reliable and valid foot classification methods to investigate the relationship between dynamic postural control and foot structure.

Subjects in our study with a pronated foot structure and subjects with a supinated foot structure had poorer standing postural control than subjects with a neutral foot structure in 2 different ways. For a laboratory or clinical standing-postural-control assessment that requires an individual to maintain the testing position for a certain period of time, with no concern for the quantity and quality of the body movements, subjects with a supinated foot structure were able to perform as well as the subjects with a neutral foot structure, but subjects with a pronated foot structure had the worst performance. For a postural-control assessment that requires not only maintaining the required test position but also minimizing body movements for a certain period of time, subjects with a pronated foot structure and subjects with a supinated foot structure demonstrated poorer performance than subjects with a neutral foot structure. Researchers and clinicians who assess standing postural-control in the barefoot condition must be aware of the potential influences of foot structure on standing postural control.

Although the results of the present study indicate that the supinated group and pronated group had poorer standing postural control than the neutral group, these findings may only generalize to individuals who have bilaterally symmetric foot structure, and whose foot structure can be defined by the criteria employed in our study. These categories may represent operational definitions of either extreme pronation or supination, compared with a rather neutral foot structure. Recruiting subjects with extreme and symmetric right and left foot structure, however, also strengthened the internal validity for finding differences in standing postural control among different weight-bearing foot structures. Another limitation for generalizing the study results is that static postural control for single-limb stance with eyes closed was evaluated with barefoot testing in the present study. The results of the study may not be applicable when individuals wear shoes or when testing is conducted with eyes open. Further studies are needed to investigate whether foot structure may result in differences in standing postural control when individuals wear shoes.

Hertel et al[13] proposed that decreased sensory afferent information resulting from reduced plantar contact area might be another possible reason why their subjects with cavus (supinated) foot structure had poorer postural control than subjects with a rectus (neutral) foot structure. This proposed mechanism, however, remains a theoretical one because plantar sensory input was not measured by Hertel et al[13] or in our study. Future studies might involve measurement of plantar contact area using pressure-sensitive pads or measurements of foot sensation for individuals with different foot structures to address this theory.

Standing postural control can be influenced by various factors, including internal neuromuscular variables or external environmental factors.[29] Although efforts were made to control intersubject differences (eg, age, body mass, body height, exercise training, and injury history), other factors, such as strength, timing of muscle activation, fatigue, history, emotion, motivation, and physical status, were not controlled and may pose a threat to the internal validity of results of the present study.

Finally, we have maintained that increased displacement and variability of the center-of-pressure position represent poorer postural stability for the tasks we tested. Previous authors have argued that increased variability is a positive human response that should be considered in relation to the dynamic system supporting posture and the postural task.[23] We have chosen traditional measures of variability of the center of pressure, as well as traditional interpretations of those variables. We recognize that standing on 1 leg with instructions to remain as motionless as possible is a dynamic task with some degree of inherent variability.

CONCLUSIONS

The results of this study indicate that subjects with a pronated foot structure and subjects with a supinated foot structure demonstrated poorer static postural control for single-limb stance than the subjects with a neutral foot structure. Different mechanisms may explain these results for each foot type. Researchers or clinicians who assess postural control in the barefoot condition must be aware of the potential influences of foot structure on postural control. Additional studies are needed to investigate the relationships among foot structure, postural con-

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trol, and the risk of lower extremity musculoskeletal injury. The results of these studies may then provide direction for clinical interventions (eg, shoe wear) that may be appropriate to improve postural control and reduce the risk of lower extremity musculoskeletal injury for individuals with abnormal foot structure.

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