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The Relationship of Body Weight and Clinical Foot and Ankle Measurements to the Heel Forces of Forward and Backward Walking

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Objective: To compare and contrast the relationships of selected static clinical measurements with the heel forces of forward and backward walking among healthy high school athletes.

Design and Setting: Single-group, cross-order-controlled, repeated-measures design. All data were collected in a high school athletic training room.

Subjects: Seventeen healthy high school student-athlete volunteers.

Measurements: We performed static clinical measurements of the foot, ankle, and knee using handheld goniometers. We used a metric ruler to assess navicular drop and a beam balance platform scale to measure body weight. Mean peak heel forces were measured using F-scan insole force sensors. Data were sampled for 3 5-second trials (50-Hz sampling rate). Mean peak heel forces were determined from 3 to 5 consecutive right foot contacts during forward and backward walking at approximately 4.02 to 4.83 km/h (2.5 to 3.0 mph). Subjects wore their own athletic shoes and alternated their initial walking direction.

Results: Forward stepwise multiple regression analyses revealed that body weight, navicular drop, and standing foot angle predicted mean peak heel forces during forward and backward walking.

Conclusions: Heel forces during forward and backward walking increase as body weight and navicular drop magnitude increase, and they decrease as standing foot angle increases. Subtle differences in foot, ankle, and knee joint postures and kinematics can affect heel forces even among normal subjects. Injury and protective bracing or taping may further affect these heel forces.

Key Words: biomechanics, gait, goniometry, posture

Lower extremity injuries are often caused by repeated, excessive, or inappropriately timed heel forces, or a combination of these, during locomotion.[1–9] Heel pain is a common malady affecting athletes with various foot and lower extremity alignments and may be related to more proximal kinetic chain dysfunction.[10] Poorly controlled rearfoot motion, decreased gastrocnemius-soleus extensibility, and hip and knee malalignments have been associated with numerous lower extremity injuries, including heel injuries.[1–10]

McPoil and Cornwall[11] examined the relationship between 17 static lower extremity measurements and rearfoot motion during forward walking among 27 healthy young adults and reported that only navicular height was predictive of maximum rearfoot pronation. The researchers concluded that, although these measurements provided important data regarding the range of motion and static alignment of the lower extremity and foot, they were limited in their overall usefulness in predicting walking stance-phase kinematics. Using multiple regression analysis, Cavanagh et al[12] examined the relationship between 27 radiographic static structural foot measurements of 50 healthy adults and dynamic foot function, reporting that soft tissue thickness and "arch-related" measurements were the strongest predictors of heel and first metatarsal head pressures (explaining approximately 35% of the dynamic plantar pressure variance). Birke et al[13] compared diabetic patients with a history of plantar surface ulceration at the first metatarsal head and matched nondiabetic controls, reporting a moderate inverse relationship between first metatarsal dorsiflexion and peak pressure at the first metatarsal head ($R^2 = -0.46$, $P < .0001$).

Backward walking and other "retro" movements are becoming increasingly popular rehabilitation methods to enhance ankle and knee joint range of motion and selectively activate muscle groups such as the ankle dorsiflexors and knee extensors.[14–16] During forward walking, weightbearing is usually initiated at the posterolateral heel, proceeds distally along the lateral foot, and terminates in the vicinity of the first metatar-

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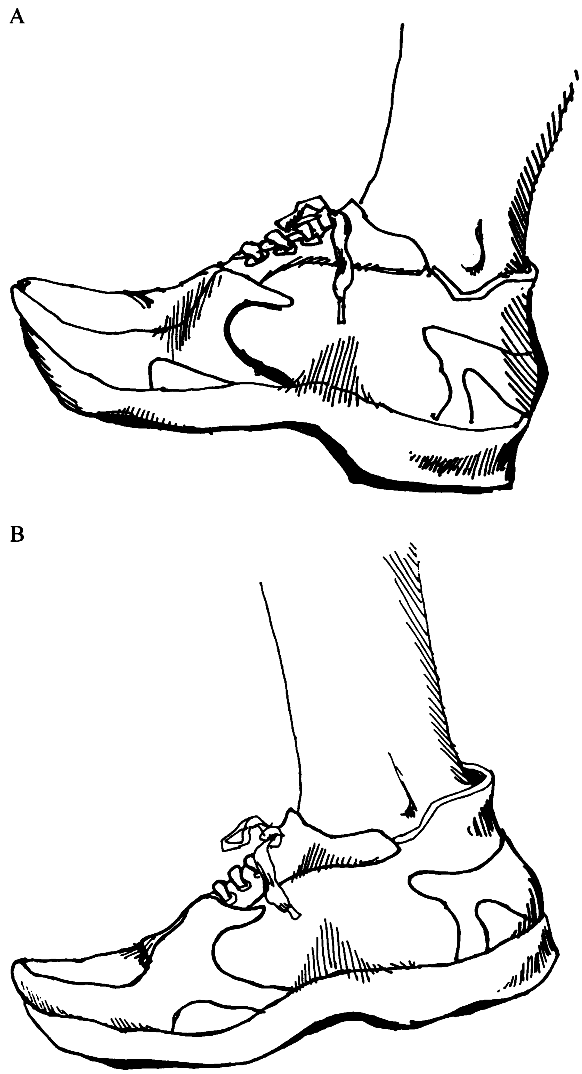

A, Forward-walking stance-phase initiation. B, Backward-walking stance-phase initiation.

sophalangeal joint.[17] During backward walking, this progression is reversed (Figure).

Based on these biomechanical considerations, altered ankle dorsiflexion, first metatarsophalangeal joint extension, or both, might affect heel forces in either walking direction. Ankle dorsiflexion increases may increase heel forces as the foot is biased toward the heel during early forward-walking stance phase or may increase heel forces during backward walking by prolonging the stance phase. First metatarsophalangeal joint extension decreases may also affect these forces as the transition from initial weightbearing to propulsion is compromised in either walking direction.

Navicular drop refers to the amount the tubercle of the tarsal navicular drops when the foot moves from neutral or balanced subtalar joint alignment to a "relaxed" subtalar joint alignment during full weightbearing stance. Neutral subtalar joint positioning was defined as the position in which the talar head was equally prominent to mediolateral palpation as the subject actively pronated and supinated the right foot during bilateral full weightbearing stance.[1] Previous studies have reported that increased navicular drop may predispose athletes to lower extremity injury.[1,11,18,19] We hypothesized that navicular drop increases would be related to increased heel forces in either walking direction.

On the basis of its bony attachments, reduced rectus femoris musculotendinous extensibility would tend to promote walking with greater hip flexion and decreased knee flexion, possibly shortening walking stride length and thereby altering heel forces. We hypothesized that increased rectus femoris extensibility would increase mean peak heel forces via greater stride lengths and more efficient walking gait.

Standing foot angle, or the Fick angle, represents the amount of toeing in or out during normal relaxed stance.[1] Reported ranges for this measurement vary with increasing age, and many studies have recorded this measurement during walking gait.[20–22] We hypothesized that increased toeing out would decrease peak heel force magnitude because more weight is accepted on the lateral aspect of the foot during walking stance phase.

Standing rearfoot angle refers to the frontal plane alignment of the posterior aspect of the heel during relaxed, full weightbearing stance.[1,23] A more valgus heel alignment is associated with greater impact force-attenuating capability and less effective propulsive-force capability.[10] A more varus heel alignment is associated with greater propulsive force-producing capability and less effective heel-contact force attenuation. Excessive or restricted motion in either direction during walking stance phase is not desired and may affect peak heel forces. We hypothesized that, as positive rearfoot alignment increased (greater valgus), mean peak heel forces would decrease.

The ability to predict peak heel forces from static clinical lower extremity measurements would be useful in screening for at-risk athletes and monitoring the effectiveness of treatment strategies designed to prevent both heel and foot injury and related proximal lower extremity microtraumatic (overuse) injury.[5–9,11,12] Although previous studies have compared static lower extremity clinical measurements with forward walking kinetics, comparisons have not been made with backward walking.[5–9] Also, previous investigations have not assessed variables such as rectus femoris and gastrocnemius-soleus musculotendinous extensibility or standing foot angle.

Arguably, many static lower extremity postures or foot, ankle, and knee soft tissue restrictions could influence heel forces during walking. Since multiple factors may influence these forces, multiple regression analysis may be the preferred method of estimating the relative effects of selected test variables. We compared and contrasted the relationships of selected static clinical lower extremity measurements with the mean peak heel forces of forward and backward walking among healthy high school athletes via multiple regression analysis.

METHODS

Subjects

Seventeen (7 female, 10 male) healthy high school student-athlete volunteers (age = 15.5 ± 1 years, weight = 65.77 ± 3.18 kg [145 ± 7 lb], and height = 170.18 ± 12.70 cm [67 ± 5 in]) participated in this study. Body weight was measured with a beam balance platform scale (Micro BioMedics Inc, Pelham Manor, NY). Subjects practiced forward and backward walking at preferred test trial velocity range (4.02 to 4.83 km/h [2.5 to 3.0 mph]) before data collection (approximated by the investigator with a handheld stopwatch).

Knee, Ankle, and First Metatarsophalangeal Joint Flexion-Extension Measurements

These measurements were collected by the principal investigator using a handheld goniometer and previously described methods.[24–26] Subjects were positioned supine and were barefoot, and the right lower extremity was used for all measurements. Pilot intratester test-retest reliability assessments (5 subjects) were performed for each measurement before data collection.

The modified Thomas test was used for hip flexor (specifically the rectus femoris muscle) extensibility assessment.[26] During this test, subjects were positioned with the right knee flexed over the edge of a treatment table. While in this position, they used their upper extremities to position and maintain the left knee against the chest. After this position was attained, they actively flexed the right knee as far as possible while maintaining right thigh-treatment table contact. When maximal active right knee flexion was reached, we took a goniometric measurement. This measurement demonstrated high intratester reliability (intraclass correlation coefficient [ICC][1,1] = .89, SEM = 2.5°).

We measured active ankle dorsiflexion with the right knee extended.[25] While in this position, subjects actively dorsiflexed the right ankle as far as possible. When maximum active ankle dorsiflexion was attained, a goniometric measurement was taken. This measurement demonstrated fair to good intratester reliability (ICC[1,1] = .81, SEM = 1.1°).

Active first metatarsophalangeal joint extension was also measured with the right knee extended.[24,25] While in this position, subjects actively dorsiflexed the right ankle to a neutral 0° dorsiflexion-plantar flexion position (as verified with a goniometer). Upon attaining this position, subjects actively extended the first metatarsophalangeal joint. When maximal active first metatarsophalangeal joint extension was attained, the primary investigator performed a goniometric measurement while stabilizing the head of the first metatarsal. This measurement demonstrated fair to good intratester reliability (ICC[1,1] = .83, SEM = 2.5°).

Subtalar Joint and Rearfoot Alignment Measurements

Before we measured rearfoot alignment and subtalar joint motion, subjects were positioned prone, while we manually applied a small mark with a felt tip marker to demarcate superior and inferior aspects of the midposterior heel and the navicular tubercle of the right foot. Goniometric measurements of rearfoot alignment and navicular drop (subtalar joint motion) were taken with the athlete in both full weightbearing relaxed and neutral subtalar joint positions. Rearfoot alignment was performed by aligning the stationary arm of the goniometer parallel with the floor and the moving arm with the heel marks. Varus rearfoot alignment was defined as negative, and valgus rearfoot alignment was defined as positive. Intratester reliability of rearfoot alignment was fair (ICC[1,1] = .76, SEM = 1.3°). Navicular drop measurements were performed using a metric ruler, and the intratester reliability was fair to good (ICC[1,1] = .84, SEM = 1.5 mm).

Standing Foot Angle Measurement

Standing foot angle was assessed while subjects assumed a relaxed, bilateral full weightbearing upright stance. We measured this variable using a handheld goniometer and the method described by Magee.[1] This measurement demonstrated fair to good intratester reliability (ICC[1,1] = .85, SEM = 2.8°).

Walking Gait Heel Force Measurements

An F-scan insole force sensor (Tekscan, Boston, MA) was inserted into the right shoe to measure peak heel forces during the stance phases of forward and backward walking. The F-scan incorporates 960 force-sensing and pressure-sensing cells beneath the entire plantar surface of the foot. The ultrathin (0.02-cm [0.007-in]) insole containing the sensors was trimmed to the subject's shoe size before insertion. Before data collection, the F-scan was calibrated to subject body weight during unilateral stance according to the manufacturer's protocol.[27] During calibration, body weights were input into the computer, after subjects assumed a unilateral right lower extremity stance position. After calibration, subjects were instructed to walk at their practice trial pace, and 3 5-second trials were collected (50-Hz sampling rate). Mean peak heel forces were determined from 3 to 5 consecutive right-foot contacts while subjects walked forward and backward at approximately 4.02 to 4.83 km/h (2.5 to 3.0 mph). Subjects wore their own athletic shoes during testing and alternated their initial walking direction to control for crossover effects.

Statistical Analysis

Descriptive statistical analysis was performed for all test variables. The following regression model was constructed to assess the predictive value of these static clinical lower extremity measurements in determining the mean peak heel forces of forward and backward walking: heel force = body weight + rearfoot alignment in subtalar neutral position + rearfoot alignment in subtalar relaxed position + navicular drop + first metatarsophalangeal joint extension + ankle dorsiflexion + modified Thomas test + standing foot angle.

Both overall (or "constrained," including all variables) and forward stepwise multiple regression analyses were performed. The forward stepwise regressions were performed to identify the predictive equation of "best fit" based on the selected independent variables.[28] Mallow's C(p) statistic was used to discriminate between regression models. This statistic considers both variance and bias in helping to select the regression model (lowest value) that best controls for overfitting or underfitting.[28] A probability level of $P < .05$ was used to indicate statistical significance. All statistical analyses were performed using SAS for Windows (version 6.11; SAS Institute, Cary, NC).

RESULTS

Descriptive statistics for static clinical lower extremity measurements are reported in Table 1. The outcomes for the multiple regression analyses of forward and backward walking are reported in Tables 2 and 3, respectively. The overall forward-walking trial multiple regression was the more predictive of the 2, with an adjusted multiple $R^2 = 0.60$, compared with an adjusted multiple $R^2 = 0.34$ for backward trials. The forward-walking regression model suggested that 60% of the variation in mean peak heel forces was explained by the selected variables. The stepwise regression for forward walking resulted in body weight, navicular drop, and standing foot angle being the best predictors of mean peak heel forces (adjusted multiple $R^2 = 0.63$, C(p) = 2.85, $P = .001$).

The overall regression model for backward walking suggested that the independent predictors explained only 34% of the mean peak heel forces, and the F value was not significant ($P = .18$). A possible reason for the better fit of the forward-walking model is the frequency with which subjects perform this task compared with backward walking, despite practicing before data collection.

When forward, stepwise regressions were performed, both the forward-walking (Table 4) and backward-walking (Table 5) regression models were significant. The stepwise regression for backward walking produced the same 3 significant variables (body weight, navicular drop, and standing foot angle) as for forward walking (adjusted multiple $R^2 = 0.47$, C(p) = 1.35, $P = .01$).

DISCUSSION

The strongest predictors for the mean peak heel forces of forward and backward walking were body weight, navicular drop, and standing foot angle. According to the forward stepwise multiple regressions for forward and backward walking, if body weight increased by 0.45 kg (1 lb), then mean peak heel forces would increase by a factor of 0.47 and 0.38, respectively. If navicular drop increased by 1 cm, then mean peak heel forces would increase by a factor of 23.7 and 18.1, respectively. If standing foot angle increased 1° in toeing-out stance, then mean peak heel forces would decrease by a factor of 1.65 and 1.38, respectively. When toeing out, less force appears to be placed at the heel and more force is placed at the lateral aspect of the foot.

The relationship between body weight and heel forces, even when walking at self-selected speeds, has tremendous relevance to the athletic arena. Since ground reaction forces may increase by a factor of 5 during running and jumping activities, heel forces should similarly increase. These increases may be particularly injurious to the heel if a rearfoot running style is employed.

Our study demonstrates the dramatic influence of subtalar joint displacement on heel impact forces during forward and backward walking. Athletic trainers can influence the magnitude and possibly the rate of this displacement via medial longitudinal arch taping, medial arch-stabilizing footwear, insoles or orthotics, and exercise programs designed to improve the neuromuscular responsiveness of ankle and subtalar joint muscles. Primary heel injury prevention can also be provided by specialized taping, which compresses fat pad tissue,[29] via padding with central relief, or heel cups.

Whereas athletic trainers can intervene at the medial longitudinal arch and the subtalar joint through both passive (taping, padding, footwear, or orthotics) and active (exercise) means, the influence of standing foot angle on heel forces may be less controllable. Based on our findings, we recommend that, in

Table 1. Descriptive Statistics for Static Clinical Measurement Variables

Variable	Mean	SD
Forward-walking peak heel force (kg)	56.36 (124.7 lb)	12.79 (28.2 lb)
Backward-walking peak heel force (kg)	54.48 (120.1 lb)	11.43 (25.2 lb)
Body weight (kg)	69.85 (154 lb)	15.20 (33.5 lb)
Standing rearfoot alignment in subtalar neutral position (°)	1.7	5
Standing rearfoot alignment in subtalar relaxed position (°)	4.8	5
Navicular drop (cm)	1.2	.50
1st metatarsophalangeal joint extension (°)	56.2	13
Ankle dorsiflexion (°)	5.4	3
Modified Thomas test (°)	102	10
Standing foot angle (°)	19.7	9

Table 2. Multiple Regression Analysis of Forward Walking: Overall Results (Dependent Variable = Mean Peak Heel Force)*

	Intercept	Body Weight	Standing Rearfoot Neutral (Valgus = +, Varus = −)	Standing Rearfoot Relaxed (Valgus = +, Varus = −)	Navicular Drop	1st Metatarsophalangeal Joint Extension	Ankle Dorsiflexion	Modified Thomas Test (Knee Flexion)	Standing Foot Angle (Toe Out = +, Toe In = −)
Regression coefficient	6.07	.481	4.07	−3.92	25.7	−.021	.075	.53	−1.48
SE	103	.20	2.64	2.87	14.2	.42	2.34	.63	.60
t score	.058	2.39	1.54	−1.36	1.80	−.05	.032	.85	−2.47

*Adjusted multiple $R^2 = .60$, F value = 3.92, $P > F = .035$.

Table 3. Multiple Regression Analysis of Backward Walking: Overall Results (Dependent Variable = Mean Peak Heel Force)*

	Intercept	Body Weight	Standing Rearfoot Neutral (Valgus = +, Varus = −)	Standing Rearfoot Relaxed (Valgus = +, Varus = −)	Navicular Drop	1st Metarsophalangeal Joint Extension	Ankle Dorsiflexion	Modified Thomas Test (Knee Flexion)	Standing Foot Angle (Toe Out = +, Toe In = −)
Regression coefficient	−.89	.42	3.27	−2.93	22.1	.27	−.57	.47	−1.22
SE	118	.22	3.02	3.2	16.8	.49	2.67	.72	.68
t score	−.008	1.84	1.08	−.892	1.35	.563	−.213	.659	−1.77

*Adjusted multiple $R^2 = .34$, F value = 2.007, $P > F = .18$.

Table 4. Multiple Regression Analysis of Forward Walking: Stepwise Results (Dependent Variable = Mean Peak Heel Force)*

	Intercept	Body Weight	Navicular Drop	Standing Foot Angle
Regression coefficient	55.8	.47	23.7	−1.65
SE	22	.13	9.39	.52

*Adjusted multiple $R^2 = .63$, C(p) = 2.85, F value = 10.1, $P > F = .001$.

Table 5. Multiple Regression Analysis of Backward Walking: Stepwise Results (Dependent Variable = Mean Peak Heel Force)*

	Intercept	Body Weight	Navicular Drop	Standing Foot Angle
Regression coefficient	66.2	.38	18.08	−1.38
SE	23.8	.14	10.1	.56

*Adjusted multiple $R^2 = .47$, C(p) = 1.35, F value = 5.74, $P > F = .01$.

addition to evaluating specific sport and position demands, athletic trainers consider body weight, navicular drop, and standing foot angle in the aggregate when deciding which athletes may be predisposed to sustaining a heel injury.

CONCLUSIONS

We found body weight, navicular drop, and standing foot-angle measurements to be the strongest predictors of mean peak heel forces during forward and backward walking. The lower adjusted multiple R^2 value and lack of statistical significance for the overall backward walking regression suggest that the subjects did not perform the task with the same consistency as forward walking or that variables other than those measured warrant consideration to accurately predict heel forces.

Even among normal subjects, subtle alterations in foot, ankle, and lower leg postures and kinematics affected mean peak heel forces during forward and backward walking. The presence of these relationships in normal subjects and probable exaggeration after injury should be of concern to athletic trainers as they treat lower extremity injuries, apply protective bracing or taping, and recommend footwear.

Similar assessments of common sport-relevant and position-relevant movements such as forward-running and backward-running directional changes (including diagonals), sudden stops and starts, and jumping may more realistically simulate the heel forces related to athletic lower extremity injuries. Further analysis should include timing and duration variables for mean peak heel forces and pressures, particularly in assessing the functional effects of taping and padding applications and other orthoses.

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