J Forensic Sci, Nov. 2002, Vol. 47, No. 6
Paper ID JFS2001244_476
Available online at: www.astm.org

Christopher M. Powers,[1] Ph.D., P.T.; Judith M. Burnfield,[2] P.T.; Phil Lim,[3] D.P.T.; John M. Brault,[4] M.S.; and James E. Flynn,[5] P.E.

Utilized Coefficient of Friction During Walking: Static Estimates Exceed Measured Values

ABSTRACT: This study compared utilized coefficient of friction (COF) measured during nonslip pedestrian gait to estimated utilized COF values calculated using anthropometric (i.e., leg length) and stride characteristic data (i.e., impact angle, step length). Twenty healthy adults walked at slow, medium, and fast speeds with kinematic and kinetic data recorded simultaneously. Estimated and measured impact angle varied with walking speed, with greater angles evident with faster speeds ($p < 0.001$ and $p < 0.05$, respectively). The estimated impact angle was greater than the measured impact angle ($p < 0.05$). Estimated and measured peak utilized COF values varied with walking speed, with higher utilized COF values evident with faster speeds ($p < 0.001$ and $p = 0.001$, respectively). Estimated utilized COF values were 86, 118, and 131% greater than measured peak utilized COF values for slow, medium, and fast speeds, respectively ($p < 0.001$). Higher estimated utilized COF values varied moderately with increased measured peak utilized COF values ($r = 0.522$; $p < 0.001$). These data suggest that impact angle and step length alone cannot be used to accurately assess the utilized COF on level walking surfaces.

KEYWORDS: forensic science, slip resistance, friction threshold, gait

Slipping is one of the most common causes of falls (1,2) and is a major concern to both industry and society owing to the associated financial costs (1,3,4). In the work environment, slipping has been identified as the primary antecedent event to falls on both stairs and level surfaces (1) and reportedly accounts for approximately 62% of underfoot accidents (2). The financial costs associated with falling are expected to exceed $85 billion during the year 2020, when it is projected that more that 17 million falls resulting in injury will occur in the United States (4).

During walking, slips result from a loss of traction (i.e., friction) between the foot and the floor. In the research setting, the utilized coefficient of friction (COF) generated during walking is determined from force plate recordings of ground reaction forces (GRFs) (5). The utilized COF is defined as the ratio between the shear (resultant of the fore-aft and medial-lateral forces) and vertical components of the GRFs generated by a person while walking across a given surface (Fig. 1).

A frequently cited theory related to the assessment of walkway slip resistance proposes that the utilized COF generated during walking can be estimated using anthropometric and stride characteristic data (6). This theory suggests that the tangent of the angle formed by the lower extremity (relative to vertical) at foot impact is equal to the ratio of shear to normal GRFs at initial contact. Ekkebus and Killey (6) derived this impact angle from gait (i.e., step length) and anthropometric data (i.e., leg length). The impact angle was defined by creating a right triangle, with the length of the lower extremity serving as the hypotenuse, and one half of the step length serving as the base of the triangle. The trigonometric sine function (i.e., opposite $\div$ hypotenuse $=$ 1/2 step length $\div$ leg length) was used to determine the angle of impact (Angle β, in Fig. 2). After estimating the impact angle, the geometric approximation of the utilized COF was calculated using the tangent of the impact angle to estimate the ratio of horizontal to vertical GRFs.

The model of Ekkebus and Killey (6) implies that as a subject increases step length, the impact angle increases and greater utilized coefficients of friction are generated (Fig. 3). As individuals typically increase walking speed through increases in stride length (7–11), static geometric calculations predict that the utilized COF during fast walking should exceed values recorded at slow walking speeds. To date, this premise has not been validated.

The purpose of this study was to compare the utilized COF as measured by a force plate during nonslip pedestrian gait to estimates of utilized COF calculated using anthropometric and gait characteristic data as proposed by Ekkebus and Killey (6). It was hypothesized that (1) estimates of utilized COF would be similar to those measured by a force plate, and (2) both the estimated and measured peak utilized COF requirements would increase with faster walking speeds.

Methods

Subjects.

Twenty healthy adults (ten females, ten males) between the ages of 23 and 37 years participated in this study (Table 1). Subjects

[1] Assistant professor, Department of Biokinesiology and Physical Therapy, University of Southern California, 1540 E. Alcazar St., CHP-155, Los Angeles, CA.
[2] Ph.D. candidate, Department of Biokinesiology and Physical Therapy, University of Southern California, 1540 E. Alcazar St., CHP-155, Los Angeles, CA.
[3] Physical therapist, Elite Physical Therapy, 11601 Wilshire Blvd., Los Angeles, CA.
[4] Senior biomechanist, Macinnis Engineering Associates, Inc., 20914 Bake Parkway, Suite 102, Lake Forest, CA.
[5] J Engineering Inc., 7636 N. Ingram Avenue, Suite 108, Fresno, CA.
Received 20 July 2001; and in revised form 29 Nov. 2001 and 9 April 2002; accepted 7 May 2002; published 18 Sept. 2002.

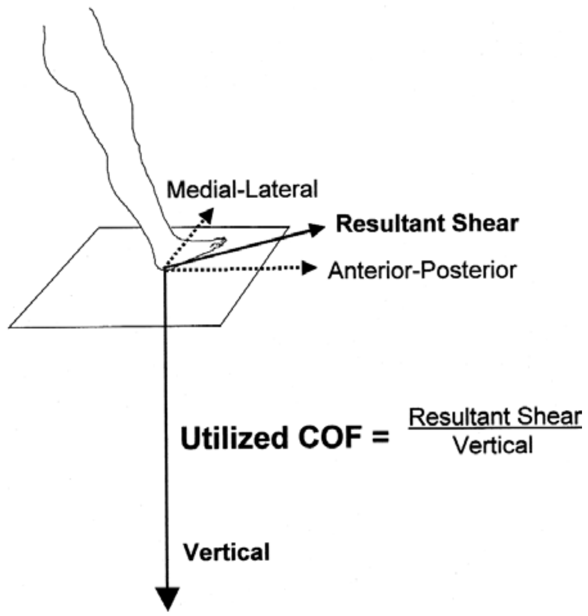

FIG. 1—*Ground reaction forces generated during walking are used to calculate the utilized COF.*

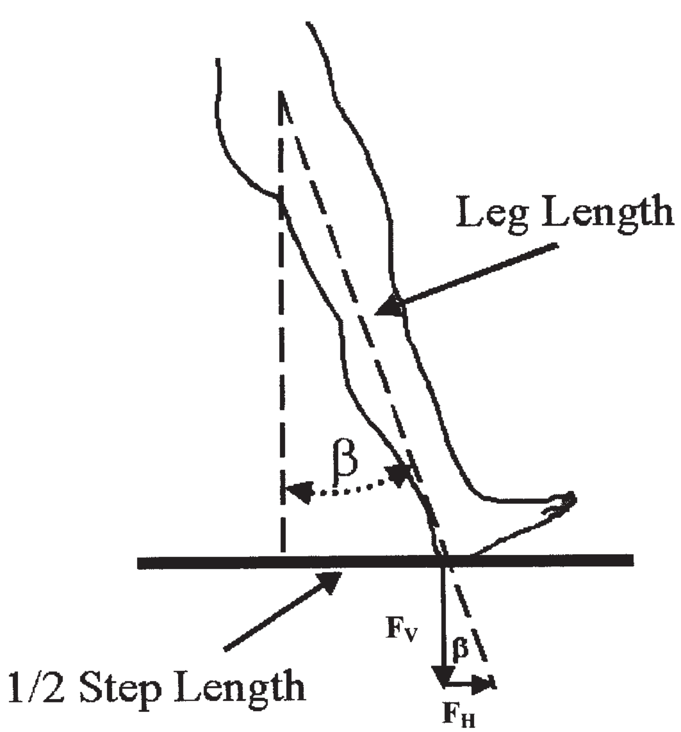

FIG. 2—*Trigonometric calculations used to determine the estimated impact angle (relative to vertical) and to estimate the utilized COF generated during walking.*

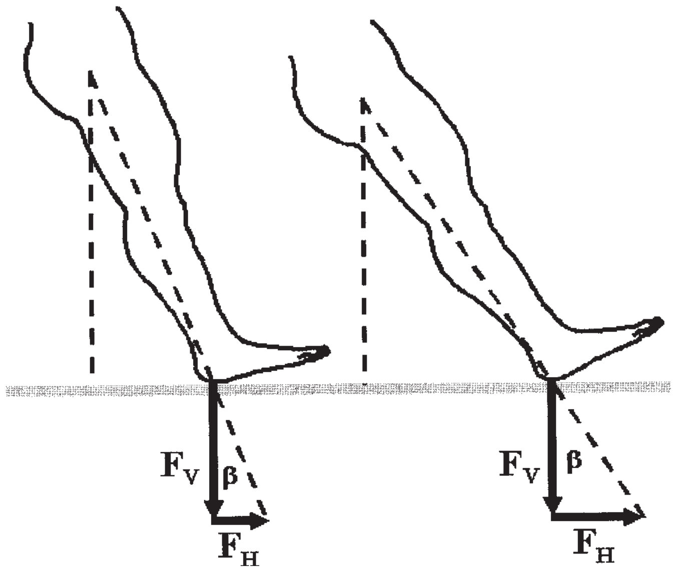

FIG. 3—*Static trigonometric estimates of the utilized COF during walking predict that a greater utilized COF is generated during long versus short strides.*

TABLE 1—*Subject characteristics, mean (SD).*

	Age, Years	Mass, kg	Height, cm	Leg Length* cm
Females	28.2 (4.8)	60.3 (5.9)	167.1 (6.5)	87.2 (3.0)
Males	28.5 (4.6)	81.5 (11.7)	177.0 (5.5)	90.8 (3.4)
Combined	28.4 (4.6)	70.9 (14.1)	172.1 (7.7)	89.0 (3.6)

* Standing measurement, right anterior superior iliac spine to medial malleolus.

were recruited from the student and faculty population at the University of Southern California (Los Angeles, CA). Only persons who were capable of independent ambulation without assistive devices were included. Subjects were excluded if they had a known history of neurologic disease or a lower extremity orthopedic condition that would interfere with walking. This was determined through a medical interview. Prior to participation, each subject was fully informed of the nature of the study and signed a human subjects consent form approved by the Institutional Review Board of the University of Southern California Health Sciences Campus.

Instrumentation

Three-dimensional motion analysis was performed using a six-camera computer-aided video (VICON, Oxford Metrics Ltd., Oxford, England) motion analysis system. Kinematic data were sampled at 60 Hz and recorded digitally on an IBM 166 MHz personal computer. Reflective markers (20-mm spheres), placed over the right fibular head and lateral malleolus, were used to determine the measured impact angle (referenced to vertical) of the lower leg at initial contact.

Ground reaction forces (vertical, fore-aft, and medial-lateral) were recorded using three AMTI force plates (Model OR6-6-1, AMTI Corp., Newton, MA) covered with smooth vinyl composition tile. These force plates were aligned in series and camouflaged within the walkway. Force plate data were sampled at 600 Hz and recorded on a 300-MHz personal computer using a 64-channel analog-to-digital converter.

Procedures

All testing was performed in the Musculoskeletal Biomechanics Research Laboratory at the University of Southern California. Prior to data collection, the length of each subject's lower extremity (right anterior superior iliac spine to right medial malleolus) was measured during standing.

Subjects walked in Oxford-style shoes (Iron-Age, Inc., Endwell, New York) that were provided for use during the walking trials. In order to determine the influence of walking speed on peak utilized COF values, predetermined target speeds were identified for slow (57 m/min), medium (87 m/min), and fast (132 m/min) walking. Subjects initially were instructed to walk at a "normal" walking speed along a 10-m walkway (dry, smooth vinyl composition tile), while looking at a spot on the wall at the far end of the walkway to avoid "targeting" a force plate. The middle 6 m of the walkway were delineated by photoelectric light switches, which were used to trigger the data acquisition computer. Walking speed was calculated following each walking trial, and only trials that were within ±5% of the targeted speed and in which a clean force plate contact occurred (i.e., the right foot contacted one of the three force plates) were accepted. All other trials were repeated. Subjects were provided with appropriate instructions to either increase, decrease, or maintain their walking speed based on a comparison of the targeted walking speed (±5%) and the recorded walking speed in combination with whether they had a clean force plate contact. Stride characteristics, kinematic and kinetic data, were recorded simultaneously.

Data Analysis

Kinematic data were analyzed using VICON 370 Workstation software (Oxford Metrics, Ltd., Oxford, England). Markers were identified manually, and three-dimensional marker coordinates were calculated. Stride length was calculated as the horizontal distance in the direction of progression of the right lateral malleolus marker from right heel contact to the next right heel contact. Step length was calculated by dividing stride length in half. The impact angle was calculated as the angle formed by the shank segment (defined by the fibular head and lateral malleolus markers) and vertical in the sagittal plane.

To duplicate methodology used by Ekkebus and Killey (6), the estimated impact angle of the entire lower extremity (relative to vertical) was calculated for each walking speed using each subject's step length and leg length as follows (see Fig. 2):

$$\text{Estimated Impact Angle } \beta_{est} = \sin^{-1} \times [(1/2 \text{ step length})/(\text{leg length})]$$

After calculating the estimated impact angle, the estimate of utilized COF for each walking condition was computed as follows (see Fig. 2):

$$\text{Estimated Utilized COF} = \tan \beta_{est}$$

Force plate data were analyzed using the VICON Workstation and Reporter software programs (Oxford Metrics, Ltd., Oxford, England). Digitally acquired fore-aft (Fy), medial-lateral (Fx), and vertical (Fz) forces were exported to ASCII file and imported to a Microsoft Office 2000 Excel spreadsheet. The Fy and Fx forces were used to calculate the resultant shear force using the following formula (see Fig. 1):

$$Fr = \sqrt{(Fx^2 + Fy^2)}$$

The utilized COF throughout stance was calculated as the ratio of Fr/Fz, and the peak utilized COF during limb loading was determined. Data were screened for spuriously high Fr/Fz ratios resulting from division by small numbers (12). On average, utilized COF values were calculated only when the Fz exceeded 59 N, which represented less than 10% of the subject's average body weight. Representative force plate and utilized COF curves for walking at medium speed are presented in Fig. 4.

Statistical Analysis

To determine if stride length, estimated impact angle, estimated utilized COF, or measured peak utilized COF varied with walking speed, separate univariate analyses of variance with repeated measures were performed. When significant differences were identified, the Tukey's Honestly Significant Difference method and Dunnett's T3 method were used to identify between which speeds differences existed. To compare the estimated impact angle and the impact angle measured by the VICON system across walking speeds, a two-way analysis of variance with repeated measures was performed. Similarly, a two-way analysis of variance with repeated measures also was performed to compare the estimated utilized COF values to those measured by force plates at each of the three walking speeds. For all two-way analyses of variance performed, if

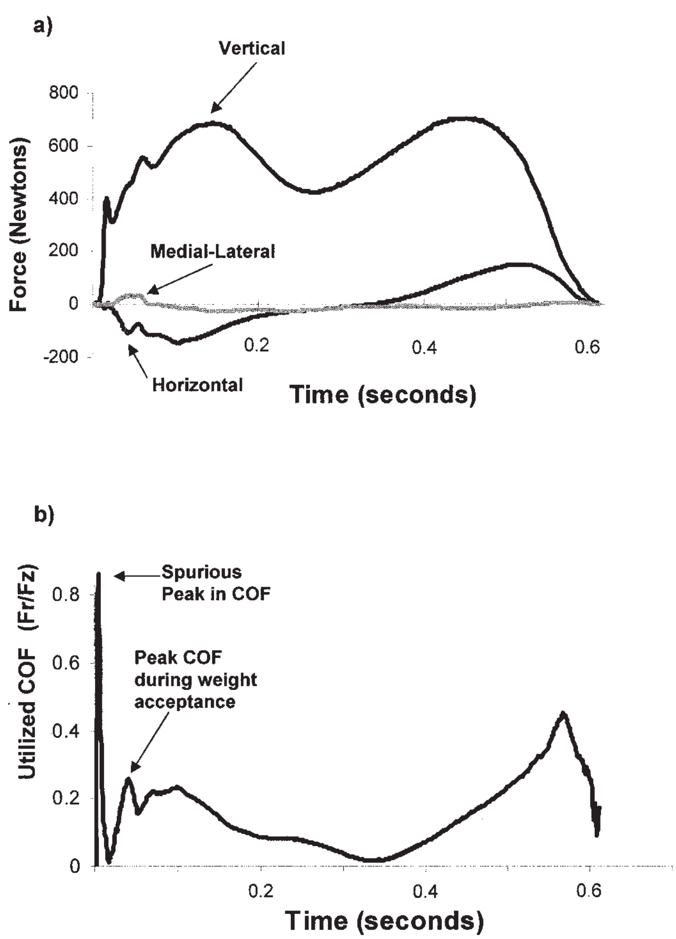

FIG. 4—*Representative tracings of* (a) *ground reaction forces, and* (b) *utilized COF during a walking trial at medium speed for a single subject. Note that the initial spuriously high spike in the utilized COF was due to a relatively low vertical ground reaction force.*

a significant interaction was found, then the main effects were considered separately.

To determine the strength of the relationship that existed between estimated utilized COF and measured peak utilized COF, a Pearson product-moment coefficient of correlation was conducted. To determine if the estimated utilized COF could be used to predict the measured peak utilized COF, linear regression analysis was conducted.

All statistics were calculated using SPSS statistical software (version 10.0; SPSS Inc., Chicago, IL). A threshold significance level of $p < 0.05$ was applied for all statistical comparisons.

Results

Stride Characteristics (Fig. 5)

The average walking speed closely approximated (within 3%) the predetermined slow, medium, and fast values. Stride length varied with walking speed, with longer strides being evident at faster walking velocities ($p < 0.001$). Post-hoc analyses revealed longer stride lengths for the fast compared to the medium speed (1.83 vs. 1.53 m) and for the medium compared to the slow speed (1.53 vs. 1.30 m) ($p < 0.001$).

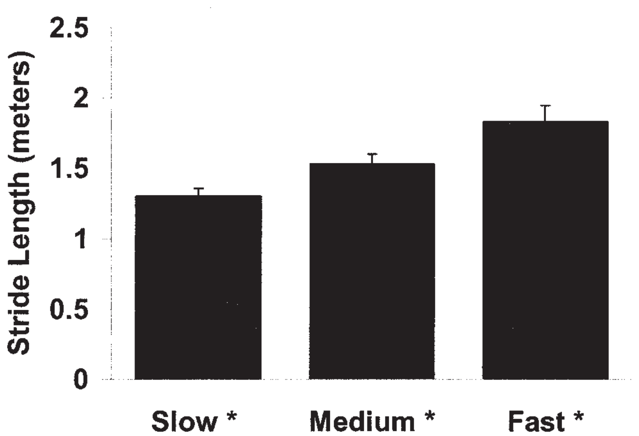

FIG. 5—*Average stride length during shod walking at slow, medium, and fast speeds (* = Fast > Medium > Slow; p < 0.001).*

Estimated and Measured Lower Extremity Impact Angle during Gait (Fig. 6)

The estimated impact angle varied with walking speed, with greater angles calculated for faster walking speeds ($p < 0.001$). Post-hoc analysis revealed a greater estimated impact angle for the fast compared to medium speed (31.1° vs. 25.5°) and for the medium compared to slow velocity (25.5° vs. 21.5°) ($p < 0.001$).

Similar to the estimated impact angle, the measured impact angle also varied with walking speed, with greater impact angles at faster walking velocities ($p < 0.05$). Post-hoc analysis revealed a greater measured impact angle for the fast compared to the slow speed (24.1° vs. 21.4°; $p < 0.04$).

The estimated impact angle was significantly greater than the measured impact angle ($p < 0.05$).

Estimated and Measured Peak Utilized COF during Gait (Fig. 7)

Estimated utilized COF values varied with walking speed, with higher estimated utilized COF values being evident with faster walking speeds ($p < 0.001$). Post-hoc analysis revealed higher estimated utilized COF values at the fast compared to medium speed ($\mu = 0.60$ vs. $\mu = 0.48$) and at the medium compared to slow speed ($\mu = 0.48$ vs. $\mu = 0.39$) ($p < 0.001$).

Similar to the estimated utilized COF values, the measured peak utilized COF values varied with walking speed, with higher utilized COF values being evident with faster walking speeds ($p = 0.001$; Fig. 7). Post-hoc analysis revealed higher measured peak utilized COF values at the fast speed (0.26) compared to both the medium ($\mu = 0.22$; $p < 0.01$) and slow walking ($\mu = 0.21$; $p = 0.001$) speeds ($p < 0.001$).

Estimated utilized COF values greatly exceeded the measured peak utilized COF values determined from force plate recordings across walking speeds.

Relationship of Estimated to Measured Peak Utilized COF during Gait (Fig. 8)

When collapsed across walking speeds, estimated utilized COF values demonstrated a moderate level of association with the measured peak utilized COF values ($r = 0.522$; $p < 0.001$; Fig. 8). The positive Pearson product-moment correlation value (r) indicated that increases in estimated utilized COF values were associated with increases in measured peak utilized COF values. Estimated

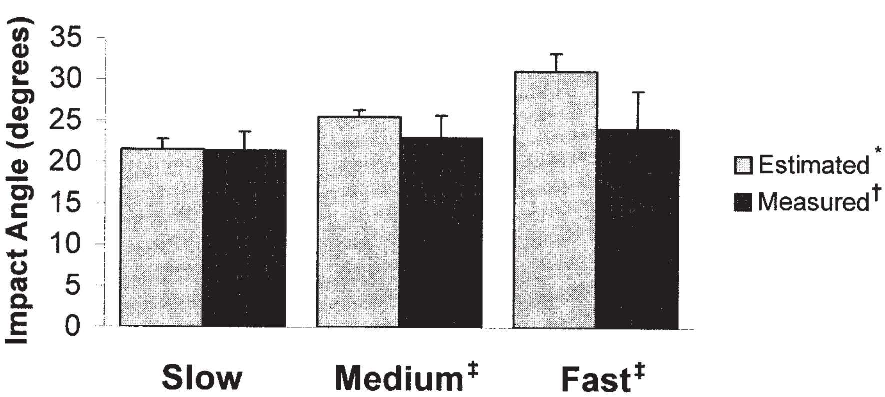

FIG. 6—*Average estimated and measured impact angle (referenced to vertical) of the right lower extremity during shod walking at slow, medium, and fast speeds. * = average estimated impact angle increased with walking speed (Fast > Medium > slow; p < 0.001). † = Average measured angle of impact increased with faster walking speeds (fast > slow; p < 0.04). ‡ = Estimated impact angles greater than measured impact angles at medium (p = 0.001) and fast (p < 0.001) walking speeds.*

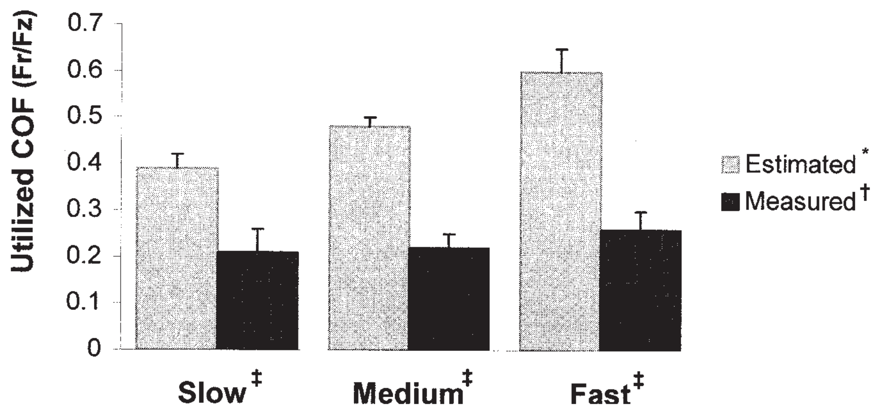

FIG. 7—*Average estimated utilized COF (tangent of the estimated impact angle) and measured peak utilized COF (Fr/Fz) during shod walking at slow (mean = 58.4 ± 1.7 m/min), medium (mean = 86.8 ± 2.6 m/min), and fast speeds (mean = 130.1 ± 3.9 m/min). * = Average estimated utilized COF increased with walking speed (fast > medium > slow; $p < 0.001$). † = Average measured peak utilized COF values were higher during the fast speed compared to the medium ($p < 0.01$) and slow ($p = 0.001$) speeds. ‡ = Average estimated utilized COF values were significantly higher than measured peak utilized COF values at fast, medium, and slow speeds ($p < 0.001$).*

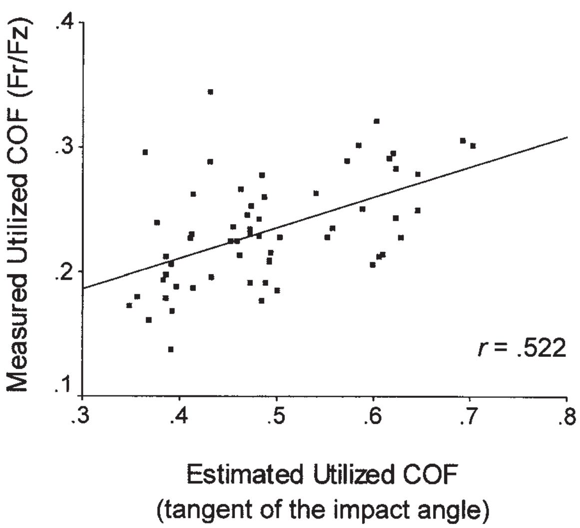

FIG. 8—*Relationship between measured peak utilized COF (Fr/Fz) and estimated utilized COF (tangent of impact angle) during shod walking at slow, medium, and fast speeds ($r = 0.522$; $R^2 = 0.273$; $p < 0.001$).*

utilized COF values accounted for 27% of the variance ($R^2 = 0.27$) observed in measured peak utilized COF values. The coefficient of determination value, R^2, reflects the amount of variance in measured peak utilized COF that could be explained based on knowledge of the estimated utilized COF values. Coefficient of determination values range from 0.00 to 1.00 with higher values being associated with a stronger accuracy of prediction.

Discussion

On average, the estimated utilized COF values greatly exceeded the utilized COF values recorded by the force plate at the slow (86%), medium (118%), and fast (131%) walking speeds. Increases in the estimated compared to measured peak utilized COF can be explained, in part, by the finding that the method used by Ekkebus and Killey (6) overestimates the angle of the lower extremity at initial contact. Ekkebus and Killey (6) assumed that the limb formed a straight line from the hip joint center to the point of impact of the foot with the ground; however, this premise is not supported by existing human kinematic data that documents that the knee is positioned in approximately 5° of flexion at initial contact (8). The error in the assumption by Ekkebus and Killey (6) is reflected by the finding that the measured impact angle was smaller than that estimated from leg length and stride length at each walking speed.

Additionally, Ekkebus and Killey (6) proposed that impact angle increased substantially with longer stride lengths. Our data, however, documented only minimal increases in the measured impact angle (3° across walking speeds) when compared to the nearly 10° increase estimated based on anthropometric and stride characteristic data. The model proposed by Ekkebus and Killey (6) assumed that increases in stride length with faster walking speeds would result in symmetrical increases in the relative angle of both the leading and trailing limb. In contrast, the current study found little change in measured impact angle despite a 41% increase in stride length between the slow and fast walking conditions. These data suggest that the observed increases in stride length were influenced by factors other than the angle of impact of the leading limb. Perry (8) suggests that the posture of the contralateral limb at initial contact has an important influence on stride length. In particular, the ability to achieve a contralateral trailing limb posture with the thigh positioned in apparent hyperextension and the heel off the ground are important contributors to stride length. Hence, subjects may be able to significantly increase stride length with only minimal increases in the impact angle of the limb with the ground.

The model of Ekkebus and Killey (6) also assumed that at initial contact, the resultant GRF vector followed a line starting at the heel and passing through the hip joint center. This implies that the leg acts as a strut when applying forces to the ground. Studies of human gait, however, indicate that immediately prior to initial contact, the body goes through a period of rapid vertical descent causing the leading limb to drop vertically by approximately 1 cm (13). This period of rapid descent results in an abrupt vertical loading of the leading limb (60% of body weight in approximately 0.02 s) (8), reducing the ratio of shear to vertical GRFs at initial contact. Additionally, studies in which the resultant vector of the shear to vertical GRF have been superimposed over a human while walking demonstrate that the vector passes anterior to the lower limb, not

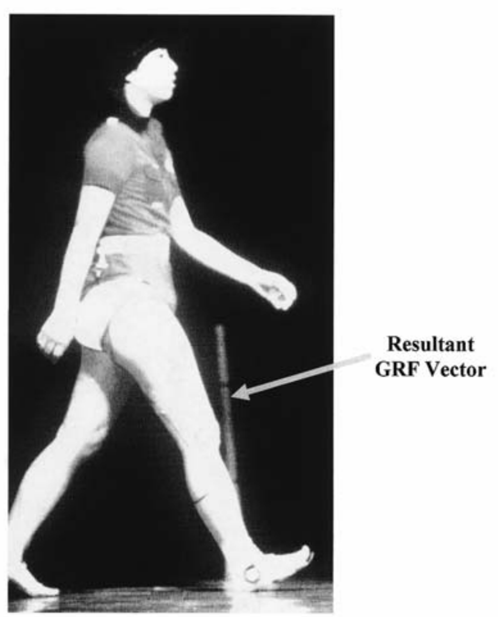

FIG. 9—*Visible resultant of the shear to vertical GRF vector superimposed over a person during walking demonstrates that the resultant vector passes anterior to the lower limb, not along it (photo printed with permission from the Pathokinesiology Laboratory, Rancho Los Amigos National Rehabilitation Center).*

through it (Fig. 9) (8). This finding is not entirely surprising as muscle activity at the hip, present prior to initial contact to decelerate forward momentum of the limb, likely alters the relative ratio of shear to vertical GRFs (8).

Results of this study have several implications. First, the model suggested by Ekkebus and Killey cannot be used to accurately assess the slip resistance required to negotiate level walking surfaces. While the moderate relationship between measured and estimated utilized COF values suggests that estimated utilized COF may serve as a basis for predicting actual utilized COF requirements during level walking, care must be taken, as the data from the current study suggest that only 27% of the variance in measured peak utilized COF values could be explained based on the estimated utilized COF values. Limitations identified in the existing model, which uses only leg length and stride length data, point to the value of developing a more comprehensive model to predict utilized COF. Selected subject-specific and condition-specific variables that might serve as predictors of utilized COF during level walking include age, gender, leg length, mass, velocity, stride length, shoe sole material, and flooring material.

Second, as the slip index scale used by the variable incidence tribometers and the portable inclineable articulated strut tribometers is based on the Ekkebus and Killey principles, care must be taken when relating instrument measurements to what would be expected by humans making contact with the ground at comparable angles.

Collectively, the findings of this study suggest that while estimated utilized COF values are related to measured peak utilized COF values during human walking, the estimated utilized COF values greatly overestimate the measured values. Dynamic influences prior to initial contact, including a period of rapid descent of the body and activity of lower extremity muscles that function to slow forward limb acceleration during swing (8), may also play an important role in modifying the utilized COF values generated during walking.

Conclusions

The findings of this study indicate that estimates of utilized COF values calculated from the anthropometric and gait characteristic data proposed by Ekkebus and Killey greatly exceed peak utilized COF values measured using a force plate. Additionally, the estimated utilized COF values calculated from the estimated impact angles demonstrated only a moderate level of association with measured peak utilized COF values across the three walking speeds. The lack of a strong association between the estimated and measured utilized COF values is likely explained, in part, by the fact that the direction of the GRF vector is not parallel to the estimated impact angle at the time of peak utilized COF.

References

1. Bentley TA, Haslam RA. Slip, trip and fall accidents occurring during the delivery of mail. Ergonomics 1998;41(12):1859–72.
2. Manning DP, Ayers IM, Jones C, Bruce M, Cohen K. The incidence of underfoot accidents during 1985 in a working population of 10,000 Meryside people. J Occup Accid 1988;(10)121–30.
3. Buck PC, Coleman VP. Slipping, tripping and falling accidents at work: a national picture. Ergonomics 1985;28(7):949–58.
4. Englander F, Hodson TJ, Terregrossa RA. Economic dimensions of slip and fall injuries. J Forensic Sci 1996;41(5):733–46.
5. Hanson JP, Redfern MS, Mazumdar M. Predicting slips and falls considering required and available friction. Ergonomics 1999;42(12):1619–33.
6. Ekkebus CF, Killey W. Measurement of safe walkway surfaces. Soaps/Cosmetics/Chemical Specialties 1973;February:40–45.
7. Andriacchi TP, Ogle JA, Galante JO. Walking speed as a basis for normal and abnormal gait measurements. J Biomech 1977;10(4):261–8.
8. Perry J. Gait analysis: normal and pathological function. Thorofare, N.J.: SLACK Inc., 1992.
9. Murray MP, Drought AB, Kory RC. Walking patterns of normal men. J Bone Joint Surg Am 1964;46(A):335–60.
10. Murray MP, Kory RC, Clarkson BH, Sepic SB. Comparison of free and fast speed walking patterns of normal men. Am J Phys Med 1966;45(1): 8–23.
11. Murray MP, Kory RC, Sepic SB. Walking patterns of normal women. Arch Phys Med Rehabil 1970;51(11):637–50.
12. Buczek FL, Banks SA. High-resolution force plate analysis of utilized slip resistance in human walking. Journal of Testing and Evaluation 1996;24(6):353–8.
13. Murray MP. Gait as a total pattern of movement. Am J Phys Med 1967;46(1):290–333.

Additional information and reprint requests:
Christopher M. Powers, Ph.D.
Assistant Professor
Department of Biokinesiology and Physical Therapy
University of Southern California
1540 E. Alcazar St., CHP-155
Los Angeles, CA 90033