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THE RELATIONSHIP BETWEEN SUBTALAR AND KNEE JOINT FUNCTION AS A POSSIBLE MECHANISM FOR RUNNING INJURIES

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ABSTRACT

The purposes of the study were: 1) to evaluate the effects of different surfaces on the relationship between subtalar and knee joint function, and 2) to examine/explore alternative approaches to the evaluation of these relationships. Five subjects ran under four different surface conditions of various hardness, while both rear and sagittal view kinematic data were collected (200Hz). Critical parameters describing the knee angle and rearfoot motion were examined in conjunction with a curve analysis technique which incorporated slope differences and curve correlations. A repeated measure ANOVA design (Surface X Subject) was used along with single subject procedures. The results of the study support a strong inter-relationship between pronation and knee joint function via tibial rotation and underlined it as a possible mechanism for injury. Moreover, discrete point analysis might not be the most appropriate methodology for evaluating dynamic functions such as rearfoot motion and knee angle. Extreme methodological care must be exercised when evaluating these functions to avoid oversmoothing and/or masking correlations and differences due to differential subject responses and individual variability. The fact that increased impact force facilitated timing discrepancies between subtalar and knee joint function resulting in a transition of the pronation curve from a unimodal to bimodal configuration, is hypothesized as a possible explanation to better understand the inter-relationships among these lower extremity functions and their relationship to running injuries.

Keywords: rearfoot motion, pronation, knee flexion, tibial rotation, running injuries, single subject

INTRODUCTION

Both rearfoot motion[1-3] and impact forces[4-7] have been implicated as primary causes of running injuries. Numerous researchers have studied the effects of shoes on rearfoot motion[3, 8-10] and speculated that excessive motion about the subtalar joint can result in various running related injuries. The impact forces associated with heel strike have also been investigated extensively[11-13] , due to their assumed association with injuries. Based upon data presented by James et al.[4], these two factors appear to contribute about equally to producing injuries; but the authors indicated that no specific anatomical structure was associated with any specific injury which does not seem unreasonable due to the numerous functional degrees of freedom within the system.

Bates et al.[11] showed that the relationship between rearfoot motion and impact can be inversely related for shoes having single density midsoles in the heel region. Soft shoes in general provided good cushioning and poor control while shoes with firm midsoles produced the opposite effects. These results have been substantiated at least in part by other investigators[3, 12, 14]. Rearfoot stability can also be influenced by shoe construction features such as heel counters/stabilizers, heel flare and dual density midsoles[14]. Considerable research on lower extremity function has focused on the effects of shoe characteristics on impact force and rearfoot stability. Little attention has been given to the effects of the running surface on these parameters. Several articles[1, 15] have documented increases in injuries resulting from harder surfaces but contain no biomechanical data. Nigg and Yeadon[16] reported comparative vertical impact data for drop tests and subject tests for 34 surfaces. Their results showed drop test differences approximately six times greater than subject differences between the lowest and the highest values. No other human data were presented nor did they indicate the correlation between the subject and drop test results. Nigg[17] suggested, however, that both types of tests are important and provide relevant information but caution must be exercised in interpreting the results.

The most prevalent site of running injuries is the knee joint[1, 4, 7]. Bates et al.[18] suggested that the mechanism for soft tissue stress about the knee might be a disruption of tibial-femoral transverse plane motion due to asynchronous timing between subtalar and knee joint actions. Pronation and knee flexion during early stance are accompanied by internal tibial rotation while supination and knee extension during late stance relate to external tibia rotation[19-21]. During the support phase of running maximum knee flexion and maximum pronation occur at approximately the same time during midstance[18]. Compensatory or over-pronation could cause a disruption in the timing pattern (maximum pronation occurring later) that could result in soft tissue stress about the knee or patellofemoral malalignment[22]. Edington et al.[23] reported that there is limited information regarding comparisons of rearfoot movement to other temporal kinematic events. They cited only two related articles from the late 70's[8, 18] and suggested that the "topic deserves much more attention than it has received to date because the effect of mistiming certain temporal events is not well understood," (p.155). More recently, Hamill et al.[24] confirmed a disruption in timing between these components resulting from a soft-midsole shoe but maximum pronation actually occurred earlier than maximum knee flexion. The authors supported the hypothesis that timing disruptions could be a mechanism for knee injury. Other researchers[9, 25] have suggested that the injury mechanism might be a function of pronation velocity but these authors have not related pronation or the velocity of pronation to knee joint function.

Van Woensel and Cavanagh[26] recognized that despite the possible implication of excessive pronation in the pathogenesis of knee injuries, only one study[8] examined the simultaneous motion of the subtalar and knee joints. Based on this observation, they examined the actions of the two joints while running on a treadmill with shoes of various midsole constructions. They found that the time to maximum pronation was an unreliable variable in describing the pattern of rearfoot motion whereas the two-phase profile using pronation/rearfoot angle and velocity was more reliable and useful. They concluded that certain subtle sagittal plane kinematic adaptations in timing and velocity patterns did occur at the knee joint in response to the shoe perturbations.

Although no human studies have substantiated the early animal study[27] that repetitive impact loading can result in joint damage, the relationship has been anecdotally established by several investigators[28-31]. Voloshin and Wosk[32] reported a relationship between reduced shock absorption across the knee joint and knee joint pathologies. The authors concluded that joint pathologies result in greater overloading of proximal joints with greater potential for degenerative injuries. Indirect evidence presented by Nigg, et al.[33] showed a relationship between hardness of playing surface and incidence of injury. In another study, Nigg[34] reported a reduction in peak tibial acceleration for softer running surfaces.

The knee is typically recognized as part of the impact absorbing mechanism during running. Less attention has been directed toward pronation as part of that mechanism. A normal amount of pronation properly sequenced is an important part of the shock absorbing system during early stance in coordination with other actions including ankle dorsi-flexion, knee flexion and hip motion[22]. Given this association along with both of the previously discussed timing considerations it would seem reasonable to incorporate selected aspects of both ankle and knee joint evaluation throughout the entire support phase when evaluating lower extremity function. The purposes of the study were: 1) to evaluate the effects of different surfaces on the relationship between subtalar and knee joint function, and 2) to examine/explore alternative approaches to the evaluation of these relationships.

METHODS

Five healthy college age male recreational runners volunteered as subjects [mass=69.9 kg (±36.04); height=182.2 cm (±8.61); age=22.8 yrs (±2.17)]. All runners performed using a heel toe footfall pattern. Prior to participation each subject signed a Subject Informed Consent Form consistent with the policy of the University of Oregon Protection of Human Subjects Committee. Data were collected for 20 trials per condition for each of four surface stiffnesses (EH=extra hard; H=hard; M=medium; S=soft). The hardest surface used was a conventional treadmill bed (Precor Model 9.4) while the three softer surfaces were achieved using an adjustable bed treadmill (Precor Model 9.3). A comfortable self selected running speed was identified for each subject running on the softest surface and this speed was maintained across all surfaces. All subjects wore the same make and model of a standard soft-soled running-type laboratory shoe during tesring sessions. Although running on the harder surfaces was physiologically easier a fixed speed was maintained so as not to confound the biomechanical parameters being investigated. Subjects ran on each surface until comfortable prior to testing. The order of presentation of the surface conditions was randomized (counter-balanced) among subjects. All data were collected during a single experimental session for each subject. Subjects were allowed to rest between conditions until they were ready to perform the next condition.

Kinematic data were collected using two 200Hz NEC high-speed video cameras interfaced to a real-time automated video based tracking system (Motion Analysis Corporation). The cameras were positioned to obtain a right sagittal and rear view of the right lower extremity during the support period. Camera distances were 14 and 15 meters, respectively and each was used in conjunction with an Augenieux Zoom lens (Model 12-200) to optimize image size while minimizing perspective error.

Reflective markers were placed on the right lower extremity to allow for path tracking[23, 25]. Sagittal view markers were placed as follows: (a) midsole of the forefoot at the head of the 5th metatarsal, (b) midsole of the heel directly below the calcaneus, (c) lateral malleolus, (d) lateral femoral condyle of the knee, (e) greater trochanter and (f) anterior superior iliac spine. Rear view markers were placed as follows: (a) center of the shoe sole, (b) upper part of the heel cap at the heel tab (both a and b markers were located so that the line between them and the horizontal formed a 90^{o} angle), (c) Achilles tendon just above the heel tab of the shoe and (d) 20 cm above marker c in the center of the leg in the standing position. The reflective markers were illuminated with a ring of light around each camera lens.

The retroreflective images from each camera were obtained and translated to planar coordinates using a Motion Analysis VP320 video-processor interfaced to an IBM compatible computer. The views were time-synchronized and time-matched by a manual switch that initiated data transmission. Twenty support phases per subject per condition were digitized. The coordinates obtained were then scaled and smoothed interactively using a Butterworth Low-Pass Filter[35]. The cut-off frequencies used were between 18 and 20 Hz for the sagittal view coordinates and 16 and 22 Hz for the rear view coordinates. The smoothed data were visually compared to the raw signal to verify the appropriateness of the processing. All files were smoothed by the same investigator to assure consistency of results. All data files were normalized to 100 points for the support period to enable mean ensemble curves to be derived for each subject-condition parameter. The coordinates from the sagittal view video data were used to identify leg and thigh positions in the anteroposterior plane (Figure 1). Subsequent knee joint angles were calculated. The vertical ankle and knee joint position coordinates were twice differentiated to obtain the vertical acceleration values for these two landmarks. The rear view coordinates were used to define the leg and calcaneus in the frontal plane and combined to define rearfoot angle which was used as an estimate of pronation.

Data were evaluated on both a group and within-subject basis. Mean vertical impact acceleration values for both the ankle and knee joints were calculated for the group as well as the individual subjects. Similarly, parameters describing critical events during the support phase (see Appendix A) were identified for each trial and mean values calculated. Mean ensemble curves were also produced for each subject-condition. In addition, mean absolute slope differences were calculated between the knee and rearfoot angle curves for consecutive pairs of data points (100 values) throughout the support period and maximum slope differences were identified. Funtionally, the slopes of the two angles represent the angular velocities that occur at the two ends of the tibia. Large differences between these velocities indicate antagonistic relationships. Finally, a curve correlation technique[36-38] was used to compare the rearfoot and knee angle curves. A high correlation indicates similar temporal characteristics between rearfoot and knee angles, while a lower correlation indicates a more asynchronous and dissimilar relationship. Furthremore, a high correlation can also be the result of coupling a unimodal knee angle curve with a bimodal rearfoot angle curve, while a lower correlation indicates that both curves are similar in shape or both unimodal. A unimodal curve is defined as a parabolic shaped curve with a single minimum, while a bimodal curve is defined as a generally parabolic shaped curve with two minimums and a local maximum in between.

The impact force values and critical event and descriptive curve parameters were all evaluated for the group using repeated measure ANOVAs (Surface X Subject) with planned comparisons between EH and the other three surfaces. Data were also evaluated using a within-subject statistical technique (Model Statistics)[39-41]. In this latter procedure, the difference between sample means is compared with the product of the mean standard deviation and a criterion test statistic based on sample size. The technique was developed to take advantage of the repeated measure concept associated with within-subject experiments rather than use an independent technique that lacks comparison sensitivity. All statistical tests were completed at the 0.05 alpha level. No adjustments were made for multiple comparisons but the number of comparisons was considered when interpreting the results.

RESULTS

Mean group data values for all surface conditions are presented in Table 1. Significant group effects were observed among the maximum vertical deceleration values for both the ankle (lateral malleolus) and knee (femoral condyle) joints. Planned comparisons ($p<0.05$) indicated greater values for the EH surface compared to the other three surfaces for both parameters. The average absolute and relative reductions in maximum impact force were 0.97 and 0.83 g's and 15.0 and 20.6% for the ankle and knee, respectively. None of the kinematic or temporal group values were significantly different although the planned comparisons indicated two significant differences (S for P2 and H for TP2) which could have been due to chance based upon the number of comparisons. These results suggest that subjects did not accommodate to the varying surface hardnesses via a neuromuscular response strategy.

In addition to group analyses, individual subject evaluations were included as part of the study. Individual subject analyses resulted in 62.2% significant comparisons overall between EH and the other three surfaces. Subject results ranged from 33.3 and 94.4% significant. Kinematic and temporal parameter results ranged from 46.7 to 80.0% and 53.3 to 66.7%, respectively. In addition, 70.0% (14 of 20) of the TK values were significantly less than the TP2 values with one value being significantly greater. Significant results were evenly distributed among surfaces with 63.3, 66.7 and 56.7% for the H, M and S surfaces, respectively, but the presence of bimodal pronation curves decreased as surface hardness decreased (5, 4, 3 and 3 for EH, H, M and S, respectively). Contrary to the group results these results suggest differential response strategies by individual subjects. To examine the relevance of the three kinematic parameters evaluated, correlation coefficients were calculated among these values for all surfaces and between these parameters and the individual deceleration values. P1 and P2 were highly correlated ($r = 0.81$) and significant relationships ($p<0.05$) were observed between P1 and both ankle ($r = 0.68$) and knee ($r = 0.69$) deceleration. The two deceleration values were also highly correlated ($r = 0.86$).

As previously indicated, one purpose of the study was to explore alternative evaluation approaches to the relationship between knee joint function and pronation. The group analysis results are given in Table 2. The mean curve correlation value is slightly greater for the softest surface which is consistent with the decreasing mean absolute slope differences and maximum negative slope differences. All of these measures indicate greater temporal similarities between the knee joint and pronation curves as surface compliance increased. These results are also consistent with the increasing number of unimodal pronation curves for the softer surfaces. Using individual subject values, the relationships among these three parameters were evaluated. The correlation coefficients for two of the three relationships were significant ($p<0.05$): curve correlation and maximum negative slope difference ($r = 0.64$) and mean absolute slope difference and maximum negative slope difference ($r = -0.70$). All three of these parameters were also related to the deceleration values with explained variances ranging from 26.0% ($r = -0.51$) to 62.4% ($r = -0.79$). Only three (33.3%) modest correlations were observed among the three discrete temporal variables and the three curve comparison techniques with an average explained variance of 41.1% ($r = 0.64$).

Based upon the individual subject analysis results additional data for the two subjects with the fewest (S1, 33.3%) and most (S4, 94.4%) discrete significant differences are presented along with an intermediate subject (S5, 50.0%) who demonstrated some extreme values (Table 3, Figures 2, 3, 4). An interesting finding from these individual subject results is the absence of any relationship between the discrete parameter data and analysis results and the within surface comparisons between the knee and pronation curves.

DISCUSSION

Considerable research in the area of running and running injuries has been completed during the past 20 years as summarized by Cavanagh[42]. However, limited efforts have been made by most researchers to relate various functional aspects as indicated by the presentation of three major topics (sagittal plane kinematics, rearfoot motion and ground reaction forces) in separate unrelated chapters. Perhaps, the reported lack of relationship between specific anatomic abnormalities and abnormal biomechanics of the lower extremity with specific injuries[22] is the consequence of this approach. It is also possible that this lack of correlation could be the result of the complexity of the human system and the associated numerous functional degrees of freedom. Another possibility, however, is that a lack of understanding of the interrelationships among the various functional components is at least part of the problem. The primary purpose of this study was to investigate selected lower extremity relationships to better understand injury mechanisms.

Potential problems with much of the research on the effects of shoe/surface characteristics on performance include a lack of statistical power and/or the use of different adaptation strategies by individual subjects[43, 44]. The lack of significant shoe effects reported[13-14] is contradictory to the anecdotal evidence in the medical/sports medicine literature which often suggests that shoes and surfaces are a cause of injury[4, 45, 46].

Subjects using different performance strategies can generate effects that cancel or compromise each other in a group design, resulting in false support for the null hypothesis even though the individual subject data values are valid and reliable[47]. The solution to this potential problem is to use a single subject design followed by a generalization, if appropriate, of the individual results. Even so, shoe/surface effects for individual subjects are generally subtle and often go undetected due to performance variability in the absence of sufficient statistical power. Since variability is an inherent part of performance even for skilled runners[48] and shoe differences are generally small, the only way to improve statistical power is to increase the number of trials per condition[39, 43]. The general approach used in this study attempted to address these potential problems since surface effects are generally similar to shoe effects.

The results from the study indicated that the differential subject response patterns were masked by the group analysis. The single subject analyses revealed numerous significant differences and provided additional insight into general performance characteristics relative to rearfoot-knee relationships. The group analysis did not provide information about how any given individual performed or might perform in the future. However, as suggested by McKenzie et al.[45] there is a need in sportsmedicine to evaluate each patient and thus provide an individual with a specific program for injury prevention or rehabilitation. This suggestion coupled with the results of the present study supports the use of single subject designs. The question of generalizability (if appropriate) of the effect on other subjects in the population can then be approached by evaluating additional subjects. It should be kept in mind, however, that increased sample size (more subjects) will not change the statistical outcome of a group analysis when subjects use different performance strategies. Also, knowing that the "average subject" transitions from a unimodal to bimodal pronation curve between conditions 2 and 3 (H and M) is of little consequence to most of the individuals regarding their own injury profile.

Most of the research related to running injuries and its association with rearfoot motion has focused on maximum values (position or velocity) and the relationships between maximum values. Two potential problems can be identified with this approach: 1) The presence of bimodal pronation curves[9, 49] could result in ambiguity between the events being evaluated especially when the two maximums are similar in value, and 2) it is probably not reasonable to think that discrete events within the support phase are the sole cause of disruptions but more likely result from a composite effect of the entire event, ie. all of stance.

Bates et al.[18] suggested that a possible mechanism responsible for various knee joint injuries to runners could be a timing discrepancy between subtalar and knee joint actions resulting in an antagonistic relationship between these joints via tibial rotation. Asynchronicity between the actions of the two joints can cause stresses at the knee joint via torques applied to the tibia at the ankle, which over a period of time could cause injury. The results of this study clearly indicate that with increased surface hardness and a corresponding increase in impact force, the rearfoot angle curve underwent a bifurcation phenomenon (Figure 3a). In addition, the data suggest that asynchronicity is more likely to occur when a bimodal rearfoot angle curve is coupled with a unimodal knee angle curve. Thus, the unimodal to bimodal transition of the pronation curve with the concurrent development of asynchronicity between the actions of pronation/supination of the foot and the flexion/extension of the knee might be associated with injurious situations.

The fact that the transition point from a unimodal to a bimodal angle curve did not occur at the same hardness/condition for all subjects is not surprising. Subjects typically enter the experimental set-up with different experiences and often respond differently. If the bifurcation of the rearfoot angle curve is an injury mechanism one would expect that the system will try to avoid the situation as long as possible. For example, one might expect an experienced runner to transition later as impact force increases than a novice runner possibly due to greater muscular strength and coordination of the necessary degrees of freedom.

The bifurcation phenomenon observed in this study can also be seen in data reported by Hamill et al.[24] while examining rearfoot motion for different midsole hardnesses. A unimodal rearfoot curve for a shoe with a 45 durometer midsole transitioned to a bimodal curve for a shoe with a 70 durometer value. Clarke et al.[9] reported two representative curves for rearfoot angle that exhibited both bimodal and unimodal characteristics, respectively. These researchers, however, suggested the unimodal curve as the model curve for rearfoot angular displacement. Other researchers[10, 25] have reported rearfoot unimodal angle curves but their results could be the result of oversmoothing. An illustration of the significance of oversmoothing for the rearfoot angle was presented by Hamill et al.[50] for data collected at a sampling frequency of 100 Hz and filtered with cut-off frequencies of 18 and 12 Hz. The bimodal nature of the curve was not present when the 12 Hz cut-off frequency was used. In both Hamill et al.[24] and the present study serious consideration was given to the smoothing process and the cutoff frequencies used to smooth the rearfoot data were similar (15-18 Hz to 16-22 Hz) in both studies.

A possible limitation of the present study was the use of a two-dimensional vs three-dimensional analysis. Hamill et al.[24, 50] reported, based on the Areblad et al.[49] study, that differences between the two types of analysis are minimal at midstance but increase as the foot moves out of plane especially during the latter portion of the stance phase (from 80% to the end of the stance phase). Therefore, Hamill et al.[50] suggested that variables such as maximum pronation, heel and leg angles and times to these events are valid for reporting rearfoot motion. Results from the present study are similar in value to those reported by others[9, 25] and the critical events (in addition to the bifurcation phenomena) occurred between 15% and 65% of the stance phase. Based on these observations, the authors do not feel that a two-dimensional analysis presents a serious limitation.

The proposed injury mechanism (asynchronicity between subtalar and knee joint actions) is also supported by the large number of observed significant differences between TP2 and K (70.0%) as well as by the increased variability of TP2 (Table 1) from the softer to the harder surface. Furthermore, the curve correlation technique[36-38] used to compare the rearfoot and knee angle curves demonstrates the dissimilarities between the two angles. As mentioned before a high correlation indicates similar temporal characteristics between pronation and knee angles, while a lower correlation indicates a more asynchronous and dissimilar relationship. Thus, the curve correlation results further emphasized the discrepancies between the continuity of the subtalar and the knee joint actions. Additionally, the use of slope differences emphasizes even more the antagonistic relationship between the flexing knee and the initial resupinating of the foot during early stance which imposes an external torque on the distal end of the tibia simultaneously with the proximal internal torque associated with knee flexion. It has been suggested in the literature that the pronation velocity is a very important variable for the evaluation of the rearfoot movement and might be associated with injury mechanisms[9, 25]. However, no previous attempt has been made to relate this variable to knee joint function. Funtionally, the slope differences of the two angles represent the angular velocities that occur at the two ends of the tibia. Large differences between these velocities represent an asynchronous symbiosis. The slope difference results show that during early stance the rearfoot velocity can be considerably greater than the knee velocity. The data from this study also suggest that a discrete point analysis might not be the most appropriate methodology for evaluating these continuous events and that the evaluation of the entire event may be more appropriate.

The combined kinematically derived ankle and knee deceleration results suggest that the adjustable low impact treadmill did accomplish its purpose of reducing the impact forces on the body. A similar treadmill was used by Elliot et al.[51] who reported increases (4.2%) in VO_2 consumption between conditions as hardness decreased. These results along with the results from the present study suggest that a more intense and injury free workout may result from training on a softer surface.

In conclusion, the results of the study support a strong inter-relationship between pronation and knee joint function. A possible mechanism for injury has been suggested and the timing discrepancy hypothesis proposed by Bates et al.[18] was supported. The results suggest that discrete point analysis might not be the most appropriate methodology for evaluating dynamic functions such as rearfoot motion and knee angle. Moreover, extreme methodological care must be exercised when evaluating these functions to avoid oversmoothing and/or masking correlations and differences due to differential subject responses and individual variability. Future research should focus on multiple perturbations to further investigate the unimodal to bimodal transition of the pronation curve with the concurrent development of asynchronicity between the actions of pronation/supination of the foot and the flexion/extension of the knee. Such a transition might be associated with injurious situations. The fact that the increased impact force facilitated the transition might be an important link to better understanding the inter-relationships among these lower extremity functions and their relationship to running injuries.

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Figure 1. Identification of joint angles.

Figure 2. Mean Ensemble curves for subject 1 conditions: a) pronation (rearfoot) and knee angle; b) pronation (rearfoot) - knee angle slope differences. Subject 1 exhibited the fewest (33.3%) significant differences in the individual subject analysis.

Figure 3. Mean Ensemble curves for subject 5 conditions: a) pronation (rearfoot) and knee angle; b) pronation (rearfoot) - knee angle slope differences. Subject 5 exhibited an intermediate (50.0%) number of significant differences in the individual subject analysis.

Figure 4. Mean Ensemble curves for subject 4 conditions: a) pronation (rearfoot) and knee angle; b) pronation (rearfoot) - knee angle slope differences. Subject 4 exhibited the most (94.4%) significant differences in the individual subject analysis.

	Surface			
Variable	EH	H	M	S
Ankle (g's)	7.5*	6.1+	5.4+	4.6+
	(2.7)	(2.9)	(2.7)	(2.3)
Knee (g's)	5.0*	3.9+	3.1+	2.5+
	(1.7)	(1.7)	(1.8)	(1.2)
P1[a] (deg)	14.4	14.7	16.5	15.4
	(3.1)	(3.4)	(1.6)	(2.1)
P2 (deg)	13.3	13.3	13.7	14.1+
	(2.3)	(2.0)	(2.0)	(2.4)
K (deg)	137.5	137.8	136.6	137.8
	(4.0)	(2.7)	(4.3)	(4.1)
TP1[a] (% stance)	22.2	24.9	17.0	17.6
	(7.4)	(5.7)	(4.7)	(7.9)
TP2 (% stance)	52.4	61.2+	53.9	51.7
	(10.6)	(13.6)	(8.7)	(2.4)
TK (% stance)	44.3	45.3	45.3	45.7
	(5.5)	(5.8)	(6.1)	(5.9)

See Appendix A for surface and variable definitions.
* Significant group ($p<0.05$)
+ Significant planned comparisons with EH ($p<0.05$)
[a] Only for cases having two maxima

Technique	Surface EH	H	M	S
Curve Correlation	0.80 (0.11)	0.78 (0.17)	0.81 (0.14)	0.85 (0.11)
Mean Absolute Slope Differences	681.0 (129.0)	618.0 (144.0)	602.0 (127.0)	577.0 (100.0)
Maximum Negative Slope Differences	-1042.0[*] (323.0)	-869.0[+] (311.0)	-897.0[+] (357.0)	-813.0[+] (253.0)

See Appendix A for surface definitions.
[*] Significant group ($p<0.05$)
[+] Significant planned comparisons with EH ($p<0.05$)

Subj.	Surface	P1	P2	TP2	TK	|AS|	|-S|	r
S1	EH	14.5	12.7	51.5	39.6+	712.0	1196.0	0.65
	H	14.2	12.6	60.0	40.6+	410.0	906.0	0.47
	M	16.2*	14.4*	49.4	40.2+	572.0	1146.0	0.59
	S	13.6	12.9	48.0	39.5+	689.0	1142.0	0.65
S4	EH	15.6	15.9	43.0	43.5	602.0	876.0	0.91
	H	----	14.9*	60.7	50.0+	602.0	694.0	0.90
	M	----	14.9*	50.8	48.8	525.0	605.0	0.88
	S	----	16.7*	50.7	48.3+	519.0	612.0	0.92
S5	EH	14.9	13.8	46.4	51.1+	743.0	1453.0	0.79
	H	16.6*	15.5*	51.7	50.5	759.0	1316.0	0.84
	M	15.1	15.0*	50.6	52.1	665.0	1312.0	0.83
	S	14.9	14.4	53.2	52.0	639.0	971.0	0.87

See Appendix A for surface and variable definitions.
* Significantly different ($p<0.05$) from EH surface
+ Significantly different ($p<0.05$) from TP2

Variable	Definition
Ankle	Ankle joint: maximum vertical impact acceleration (g's)
Knee	Knee joint: maximum vertical impact acceleration (g's)
P1	First maximum rearfoot angle (deg)
P2	Second maximum rearfoot angle (deg)
K	Minimum knee joint angle (deg)
TP1	Time to P1 relative to stance (%)
TP2	Time to P2 relative to stance (%)
TK	Time to K relative to stance (%)
|AS|	Mean absolute slope differences between the knee and the rearfoot angle curves (deg/sec)
|-S|	Absolute value of maximum negative slope difference (deg/sec)
EH	Surface condition: Extra hard
H	Surface condition: Hard
M	Surface condition: Medium
S	Surface condition: Soft