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ABSTRACT

Paragraph Number 1 **Purpose:** It has been suggested that during running proper coordination between subtalar joint pronation/supination and knee joint flexion/extension via tibial rotation is important to attenuate ground reaction impact forces (GRIF). Lack of coordination may produce over time a wide range of injuries. The goal of this study was to investigate the relationship between subtalar pronation/supination and knee flexion/extension with GRIF increases during distance running. **Methods:** Ten subjects ran under different speeds (a self-selected pace, 10% faster, 10% slower and 20% faster), and over different obstacle heights (5%, 10%, 15% of their standing height) on their self-selected pace. Sagittal, rear view kinematic data and GRIF data were collected. The biomechanical results were also compared with data from a clinical evaluation of the subjects. **Results:** Speed changes and obstacle heights produced increases in GRIF and differences between rearfoot and knee angular velocities. The higher the obstacle and the faster the speed, the greater the GRIF and the greater the velocity differences. A change of the rearfoot angle curve from a unimodal (one minimum) to a bimodal (two minimums) parabolic configuration was also observed. The appearance of the second minimum was attributed to a lateral deviation of the tibia as a rebound effect due to the increased impact with the ground. The velocity differences between the actions of the subtalar and the knee joint, which in essence capture the antagonistic nature of their relationship, produced the highest correlation with the clinical evaluation. **Conclusions:** It was suggested that a possible mechanism responsible for various running injuries could be lack of coordination between subtalar and knee joint actions. This mechanism may have potential for predicting runners with susceptibility to injury.

Keywords: SUBTALAR PRONATION SUPINATION, KNEE FLEXION EXTENSION, OBSTACLE HEIGHTS, SPEED CHANGES, COORDINATION, RUNNING INJURIES

INTRODUCTION

Paragraph Number 2 Running, as one of the most popular forms of exercise and recreation, can address many of the objectives published in the "Healthy People 2000" report (24). It can be used to provide "health benefits for all people" since it is very inexpensive (requires only a pair of shoes and space) and can dramatically improve cardiovascular, musculoskeletal and mental fitness to foster "health promotion and disease prevention". As indicated by the Surgeon General Report on Physical Activity and Health (25), the weight-bearing nature of running can increase bone strength and delay the onset of frailty as people age and protect against disorders such as osteoporosis. However, running injury-free and gaining all these positive attributes is very difficult. Various studies estimate the yearly incidence of injuries among runners to be 37-56% (26). The majority of these injuries are overuse lower extremity injuries (70-80%; 17, 26) and they are located at the knee and below for about 80% of all injuries (17, 26). To prevent and treat such injuries it is essential to understand the mechanisms that predispose and lead to injury.

Paragraph Number 3 The medical community reports that there has not been any significant progress made in our understanding of the mechanisms of running injuries (14, 17, 22, 26). According to various estimates, fewer than 20% of treatments used in sports medicine are supported by solid scientific evidence (22). Diagnosis, treatment, and prevention of running injuries is reported to be more of a speculative art than a science and the exchange of information between biomechanics and clinical medicine has been limited (14, 17, 22, 26). Based upon an extensive review, van Mechelen (26) identified four factors significantly related to running injuries: a) previous injury or injury proneness, b) lack of experience, c) running to compete, and d) excessive weekly running distance. Malalignment, restricted range of motion, stability of running pattern, surface hardness, shoes, and orthotics are considered related factors but support for them is either scanty or not clear (26). It was further suggested that static malalignment of the lower extremity may not be as significant as a dynamic, functional deficit in predisposing an athlete to stress injury (26).

Paragraph Number 4 During running the repetitive and violent encounter between the foot and the ground results in impact forces that have to be absorbed by the supportive structures. The joint actions of knee flexion and subtalar joint pronation are both associated with attenuation of these impulsive shock loads (2, 4, 15, 29). Specifically, Kim et al. (15) reported that 70-80% of impact peak reduction could be credited to the knee joint. At the distal end of the tibia, subtalar pronation allows for the impact forces to be absorbed during a longer period by the supporting structures reducing the effective magnitudes of these forces. Without subtalar pronation these forces would have to be abruptly and directly absorbed by the supporting structures, causing problems associated with excessive stress (2, 4, 11, 21).

Paragraph Number 5 A mechanism that have received limited attention in the biomechanical research is the coordinative action between pronation of the subtalar joint and knee motion. In the tibiofemoral joint, the medial condyle extends further distally (1.7 cm) than the lateral condyle (19). This added length causes the tibia to rotate around the femur during flexion and extension. Specifically, as the knee flexes and then extends, the tibia descends and then ascends the curves of the medial femoral condyle and simultaneously rotates externally. This motion is reversed as the tibia moves back into flexion (19). At the other end of the tibia, the subtalar joint has been modeled as a mitered hinge joint (11, 21), where rotation of the foot about the longitudinal axis is transmitted to the tibia, imposing rotation on it about its long axis. Thus, rotation of the foot from a supinated to a pronated position results in an internal rotation of the tibia through its longitudinal axis. The opposite is also true (11, 21). In running and during early stance, the subtalar joint pronates and the knee joint flexes, allowing for the impact forces to be absorbed. As described above, these joint actions are accompanied by an internal tibial rotation. On the contrary and during late stance, the subtalar joint supinates and the knee joint extends. These actions are accompanied by external tibial rotation. The transition between knee joint flexion to extension and subtalar joint pronation to supination occur at approximately the same time during midstance (2, 5, 7). However, if the subtalar joint begins the transition at a different time than the knee joint, an antagonistic relationship may be present. Such a relationship could result in soft tissue stress and, with multiple repetitions as is the case in running, may lead to an injury.

Paragraph Number 6 However, limited research exist that examined the coordinative actions between the two joints and their possible association with specific injuries. Torsional abnormalities are considered to be a possible cause of osteoarthritis at the knee (30). Specifically, reduction in external tibial rotation has been associated with osteoarthritis (30). Furthermore, abnormal or increased pronation may also lead to injuries in diverse locations of the lower extremities. Messier and Pittala (18) found that a control group of non-injured runners exhibited less pronation than three injury groups. However, as Edington et al. (7) reported, there is limited information comparing the rearfoot motion to other timing events. They cited only two related articles from the late 1970's (2, 3) and suggested that the topic deserves much more attention. Recently, Hamill et al. (9) reported timing differences between subtalar and knee joint actions resulting from a soft shoe condition with maximum pronation occurring before maximum knee flexion. Stergiou and Bates (23) observed that increases in impact forces due to increased surface hardness can cause a change from a unimodal to a bimodal configuration for the rearfoot angle curve. The rearfoot angle describes the movements of pronation/supination of the subtalar joint. A unimodal rearfoot curve is defined as a parabolic curve with a single minimum, while a bimodal rearfoot curve exhibits two minimums with a local maximum in between. Stergiou and Bates (23) suggested that lack of coordination between the subtalar and knee joint actions is more likely to occur when a bimodal rearfoot angle curve is coupled with the typical unimodal knee angle curve. They also suggested that increases in impact forces may augment the abnormal timing between the two joint actions and disrupt their coordinative motions which are important for the shock absorption mechanism. They recommended that abnormal timing may be a mechanism for running injuries. However, they suggested that the phenomenon requires further investigation via multiple perturbations and future research should include clinical evaluations to support its prevalence and thus, its external validity.

Paragraph Number 7 The purpose of this study was to further investigate the relationships between a) frontal plane kinematics (subtalar pronation/supination), b) sagittal plane kinematics (knee flexion/extension) and c) kinetics (ground reaction impact forces) during distance running to gain a better understanding of lower extremity function and underlying mechanisms which might be implicated as possible precursors to injury. To accomplish this purpose the following hypothesis was developed: as ground reaction impact forces increase the actions of pronation/supination of the subtalar joint and flexion/extension of the knee joint will become more asynchronous and at some point the rearfoot angle curve will change from a unimodal to a bimodal configuration. To test this hypothesis, subjects ran under different conditions (with different speeds and over obstacles of different heights) to facilitate increases in ground reaction impact forces. Subsequently, the knee angle during support and parameters describing pronation were examined in conjunction with the changing impact force values.

Paragraph Number 8 In addition, as Messier and Pittala (18) have mentioned biomechanical studies are sometimes just descriptive and as a result they fail to establish external validity. Conversely, sports medicine publications are often based upon opinions, without scientific data to support them. Thus, as a secondary purpose and in an effort to increase the clinical external validity of the results of this study, subjects' rankings for susceptibility to injury were constructed from both a clinical evaluation and the biomechanical data. Finally, the clinical rankings were correlated with the biomechanical rankings to determine possible relationships.

METHODS

Subjects

Paragraph Number 9 The subjects of this investigation were eight healthy male (n = 4) and female (n = 4) runners who had been running a minimum of 10 miles per week for at least 1 yr (mean age: 25.9 yr; mean body mass: 73.45 kg; mean height: 177 cm). Prior to testing, each subject read and signed an informed consent form consistent with university and American College of Sports Medicine policy.

Instrumentation

Paragraph Number 10 An Advanced Medical Technologies Inc. (AMTI Model OR6-5-1) force platform system was used to collect the ground reaction force data. The force platform was installed in the middle of a 30 m runway in the Biomechanics Laboratory at the University of Oregon. The top surface of the platform was flush with the laboratory floor to create a complete and safe running surface. The platform was mounted on a stainless steel base plate secured to a concrete block that was an integral part of the building structure. An AMTI signal conditioner/amplifier (Model SGA6-3) was employed in conjunction with the force platform. The signal conditioner/amplifier was interfaced with an Ariel Performance Analysis System (APAS) containing a 32 channel analog to digital sampling module interfaced to a 80386-processor computer. Three channels of force signals (Fz, Fy, and Fx orthogonal components) and one synchronizing channel, were collected at a sampling rate of 1000 Hz.

Paragraph Number 11 Kinematic data were collected at a sampling rate of 200 Hz using two 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 (frontal plane) view of the right lower extremity during the support period. Camera distances were 14 and 11 meters, respectively and each was used in conjunction with an Augenieux Zoom Type 10x12A lens to optimize image size while minimizing perspective error. Prior to recording the movement, reflective markers were placed on the subject’s right lower extremity to allow for path tracking. Specifically, the sagittal view markers were placed as follows: a) lateral malleolus, b) knee joint center, and c) greater trochanter. Rear view markers were placed based on the absolute approach (5, 7) and as follows: a) midline of the Achilles tendon between the lateral and medial malleolus, b) below the belly of the gastrocnemius on a line joining the previous marker with the bisection of the leg at the level of the popliteal fossa, and c) with d) on the heel counter of the shoe and on a line that approximates the bisection of the posterior aspect of the calcaneus. The reflective markers were illuminated with a 650W lamp from a Mini-Brute 9 fixture (Berkey-Colotran Model 112-001) located directly above the camera lens. The retroreflective images from each camera were obtained and translated to cartesian coordinates using a Motion Analysis VP320 video processor interfaced to an 80486 processor computer.

Paragraph Number 12 Data collection by the APAS and the video tracking system was triggered by a manual transistor/transistor/logic (TTL) switch. The two video views were time-synchronized by the TTL switch that initiated data transmission. In addition, two light emitting diodes (LED) were used to synchronize the video and force data. The Fz channel was connected in parallel to the LEDs. Upon foot contact a threshold analog signal from the Fz channel triggered the LEDs and also generated a voltage pulse square wave that was sampled by the APAS. One LED was included in each field of view of the two cameras. The initial frame where the light appeared in each camera and the time of initiation of the square wave were used to synchronize the video and force data and identify foot contact.

Protocol

Paragraph Number 13 Subjects attended three test sessions on three different days. On the first day, subjects underwent an clinical examination by a medical doctor (Stanley L. James). The physician administered a battery of clinical tests to evaluate lower extremity alignment and geometry (Appendix A). The specific methods used to conduct these clinical tests are described in detail elsewhere in the literature (13, 16, 28). Information regarding subjects' injury history and running experience was also gathered. On the second day the subjects ran at different speeds and on the third day they ran over obstacles of different heights.

Paragraph Number 14 On both obstacle and speed test sessions, the subjects were given time to accommodate to the experimental set up and to adequately warm-up prior to testing. Warm-up consisted of running through the testing area without concern for stepping on a force platform. During warm-up the subject established a comfortable running pace which was recorded. The subjects' running speed was monitored over a 3 m interval using a photoelectronic timing system (Lafayette Performance Pack model 63520). This speed $\pm$5% was used as the baseline self-selected pace speed for subsequent testing. Following this procedure a foot placement marker was located approximately 10 m before the timed interval to allow for a normal right foot contact on the force platform. Each trial consisted of a run of approximately 40 m. Data transfer from the cameras to the computer and the qualitative inspection of the force curves allowed for a 1 min inter-trial rest interval.

Paragraph Number 15 For the speed test session, the subjects ran at four different speeds: the comfortable self-selected pace, 10% faster, 10% slower and 20% faster. For example, a subject that identified 3 $m \bullet s^{-1}$ as the comfortable self-selected pace also performed at 3.3, 2.7 and 3.6 $m \bullet s^{-1}$. Each speed condition consisted of ten trials and the order of presentation of the conditions was randomized. For the obstacle test session, the subjects ran at their previously established comfortable self-selected pace over obstacles of three different heights: 5%, 10%, and 15% of their standing height. For example, a subject who was 1.83 m tall ran over obstacles of 9.15, 18.3 and 27.45 cm. By comparison a typical curbstone separating cars in a parking lot, or the height of a sidewalk above the street are typically 15 to 17.5 cm. The obstacles were placed directly before the force platform so that the subject had to clear the obstacle with the right leg and land on the force platform. While the subjects were performing at their self-selected pace, a marker was positioned one step before the force platform to identify left foot position. When the obstacle was placed on the runway the subjects were instructed to hit the marker with their left foot prior to clearing the obstacle with the right leg. Using this procedure insured that the subject did not change their stride length when clearing the obstacle. The subjects were also instructed to run over the obstacles and avoid jumping over them, in order to maintain a normal heel-toe running pattern. The obstacles were made of extremely light wood so that if a subject stepped on or hit the obstacle by mistake while running, the obstacle was destroyed. This minimized the risk of the subject's fear of tripping and/or falling. Each obstacle condition consisted of ten trials and the order of presentation of the conditions was randomized.

Paragraph Number 16 Both the obstacle heights and the different speeds were established upon pilot work and the corresponding running literature (9). The information evaluated suggested that with these conditions adequate statistical power (greater than 70%) could be achieved to detect significant differences of 2.5-3 $N \bullet kg^{-1}$ in ground reaction impact force values (normalized to body mass) using eight subjects for ten trials per condition.

Data Reduction

Paragraph Number 17 The first peak value (impact peak) of the vertical ground reaction force (Fz) was identified for each trial via laboratory software. Via this software the user can inspect a plot and then with the help of a superimposed cursor can designate two points on the plot. An algorithm can then be used to select the maximum value between the designated points. Subsequently, the ground reaction impact force (GRIF) values were normalized for body mass, and mean values were calculated across trials for each condition of each subject. The group means of all subjects were also calculated for each condition.

Paragraph Number 18 The kinematic coordinates were scaled and smoothed using a Butterworth Low-Pass Filter with a selective cut-off algorithm based on Jackson (12). The cut-off frequency values used were 13-16 Hz for the sagittal view coordinates and 16-20 Hz for the rear view coordinates. From the scaled and smoothed coordinates the following angles were calculated (Figure 1): a) from the sagittal view the knee angle, and b) from the rear view, the angles of the foot and the leg with respect to the left horizontal. Then, the rearfoot angle is calculated by subtracting the leg angle from the foot angle. Subsequently, the angular position data were

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differentiated using a cubic spline routine to calculate angular velocities. All kinematic parameter data files were normalized to 100 points for the support period using a cubic spline routine to enable mean ensemble curves to be derived for each condition of each subject.

Data Analysis

Discrete Point Analysis

Paragraph Number 19 The minimum knee angle, the maximum negative rearfoot angle, and their respective times of occurrence were also identified for each trial. Functionally, the minimum knee angle corresponds to maximum flexion of the knee joint and the maximum negative rearfoot angle corresponds to maximum pronation of the subtalar joint. Thus, their respective times of occurrence were named as time to maximum knee flexion (TMKF) and time to maximum pronation (TMP). In addition, the absolute differences between TMP and TMKF (|TMP−TMKF|) were also identified for each trial. The mean values of TMP, TMKF, and |TMP−TMKF| were calculated across trials for each condition of each subject. Group means of all subjects were also calculated for each condition. These critical event parameters were used to evaluate timing differences between the actions of the two joints.

Analysis of the entire support period.

Paragraph Number 20 To examine the actions of subtalar pronation/supination and knee flexion/extension over the entire support period additional techniques were employed. These techniques were a) curve correlations, and b) velocity differences.

Paragraph Number 21 The curve correlation technique was introduced by Derrick et al. (6). Based on this technique a point by point Pearson Product moment correlation coefficient was calculated between the corresponding data points from the rearfoot and knee angle mean ensemble curves. A high correlation indicated similar angle curves and thus, proper coordination between the actions of the subtalar and the knee. A lower correlation revealed the opposite. A lower correlation could also be the result of coupling a unimodal knee angle with a bimodal rearfoot angle curve, while a higher correlation indicated that both curves were similar or both unimodal. A unimodal curve was defined as a parabolic curve with a single minimum, while a bimodal curve exhibited two minimums with a local maximum in between. Group means of all subjects and for each condition were also calculated from the curve correlation values.

Paragraph Number 22 In the velocity differences technique (23), a point by point difference was calculated between the corresponding data points from the rearfoot and knee angular velocity mean ensemble curves. By this technique, a new curve was generated that represented the angular velocity differences throughout the support period. Functionally, large differences between the velocities indicated antagonistic relationships at the two ends of the tibia and thus, possible injurious situations. To describe the generated velocity differences curve by a single number, the mean of the absolute values of the curve was calculated. This mean of the absolute values captured the entire curve throughout the support period. The greater this number the larger the velocity differences, and vice versa. Group means of all subjects and for each condition were also calculated from the mean absolute velocity differences (MAVD) values.

Construction of rankings

Paragraph Number 23 To increase the clinical external validity of the results of this study, subjects' rankings for susceptibility to injury were constructed from both the clinical evaluation and the biomechanical data. The biomechanical subjects' rankings were based on the variable ability of every subject to sustain proper coordination through the changes introduced in the experiment (obstacle heights and speed changes). Proper coordination was defined as the ability of a subject to maintain time matching and sequencing between the actions of subtalar pronation/supination and knee flexion/extension. Lack of coordination was assumed to be an injurious situation. Therefore, subjects' rankings were constructed based on two different methods to examine coordination from the biomechanical data. These methods were a) curve correlations and b) velocity differences.

Paragraph Number 24 The curve correlations (CC) ranking was constructed as follows. After the rearfoot and knee angle mean ensemble curves were correlated, the CC values were averaged across conditions and across test sessions for each subject. These means and their respective standard deviations were used to rank the subjects for susceptibility to injury. One subject's rank was constructed based on the means. A high mean indicated high values for CC and similar angle curves across conditions, while a low mean presumed the opposite. Similar angle curves was interpreted as maintaining time matching and sequencing of events (flexion/extension and pronation/supination) throughout the support period. The subject with the highest mean was considered as least likely to sustain an injury under varied running conditions, because this subject maintained coordination and was assigned a rank of one. The subjects were also ranked based on the standard deviations. For this ranking, a low standard deviation indicated small variations among the CC values across conditions and few dissimilarities between the two angle curves, while a large value presumes the opposite. The subject with the lowest standard deviation was considered as least likely to sustain an injury under varied running conditions and was assigned a rank of one. The subjects' scores from the two rankings (means and standard deviations) were added and the sums were used to construct a final CC ranking. For this ranking, the subject with the lowest score was assigned a rank of one, and vice versa.

Paragraph Number 25 The velocity differences (VD) ranking was constructed as follows. The MAVD values were averaged across conditions and across test sessions for each subject. These latest means and their respective standard deviations were used to rank the subjects for susceptibility to injury. One subject's rank was constructed based on the means. A low mean value indicated smaller differences between the two angular velocities across conditions, while a large value presumes the opposite. Large velocity differences were interpreted as antagonistic relationships at the two ends of the tibia and thus, as lack of proper coordination. The subject with the lowest mean was considered as least likely to sustain an injury under varied running conditions because this subject maintained coordination (small velocity differences) between the two actions. This subject was assigned a rank of one. The subjects were also ranked based on the standard deviations. For this ranking, a low standard deviation indicated small variations between the angular velocity differences across conditions, while a large value presumes the opposite. Thus, the subject with the lowest standard deviation was considered as least likely to sustain an injury under varied running conditions and was assigned a rank of one. The subjects' scores from the two rankings (means and standard deviations) were added and the sums were used to construct a final VD ranking. For this ranking, the subject with the lowest score was assigned a rank of one, and vice versa.

Clinical evaluation and ranking

Paragraph Number 26 A major problem that has delayed the prevention of running injuries is the absence of clinical evaluation to validate biomechanical results. Thus as a secondary purpose of the present study and in an effort to increase the external validity, subjects' rankings for susceptibility were constructed based on an clinical evaluation and compared to the biomechanical results. Stanley L. James MD FACSM, an orthopedist and sports medicine specialist conducted the clinical examination. Additionally and in an independent evaluation, Dr. James ranked the subjects on susceptibility to injury based on injury history

(IH), running experience (RE) and the clinical examination (OE). The subjects' ranking was based on the clinical experience of Dr. James over 35 years. This method is not based on any previous research and it is subjective. However, if the evaluation of patients in the clinics is currently based on the same criteria (self-perception of symptoms' importance), then it seems logical to attempt to validate the biomechanical results with the clinical evaluation of an experienced orthopedist. Dr. James, a well published (2, 3, 13, 14) orthopedist and sports medicine specialist who has treated runners for the last 35 years, was sought to provide the clinical evaluation.

Statistical Analysis

Paragraph Number 27 One-way repeated measures ANOVA (condition by subjects) were performed on the subject means for each test session (obstacle heights and speed changes) for GRIF, TMP, TMKF, |TMP-TMKF|, and MAVD. In tests that resulted in a significant F-ratio ($P<0.05$), a Tukey multiple comparison test was performed to identify the location of the significant differences. In addition, Spearman rank order correlations were used to compare the clinical subjects' rankings for susceptibility to injury with the biomechanical subjects' rankings. All statistical tests were completed at the 0.05 alpha level.

RESULTS

Paragraph Number 28 The group analysis results are presented in Table 1. The GRIF results were statistically different for both test sessions. The post-hoc analysis revealed that from all the possible comparisons for the speed session, only the 20% faster condition was statistically different from both the 10% slower condition and the self-selected pace condition. For the obstacle session, all possible comparisons were statistically different. These results indicated that the experimental design was successful in facilitating significant increases in impact forces between conditions, with the obstacle conditions producing the more consistent effects. Furthermore, the MAVD results were statistically different for both sessions. The post-hoc analysis revealed that four out of six possible comparisons were statistically different for the speed session. For the obstacle session, the 15% obstacle condition was statistically different from both the 5% obstacle and the self-selected (no obstacle) condition. These results indicated that the speed experimental protocol produced more consistent changes for the velocity differences between the two joint actions, while for the obstacle experimental protocol only the highest obstacle was successful in producing significant effects. For the rest of the group analysis comparisons, only the TMP results were statistically different for the obstacle conditions. The post-hoc analysis revealed that the self-selected pace (no obstacle) condition was statistically different from both the 5% and 15% obstacle conditions. No significant differences were found for TMKF and |TMP-TMKF|.

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Paragraph Number 29 It is observed from Table 1 that the group curve correlation coefficients between the rearfoot and the knee angle mean ensemble curves exhibited an inverse relationship with the group GRIF results for the obstacle conditions (r = −0.96 with 92% explained variance). The curve correlation values decreased from the self-selected pace (no obstacle) condition to the 15% obstacle condition, which is opposite from the group GRIF values. Such a relationship is not observed for the speed conditions. However, the group mean absolute velocity difference (MAVD) results exhibited a direct relationship with the group GRIF results for both test sessions (r = 0.93 with 86% explained variance and r = 0.91 with 83% explained variance for the obstacle and speed conditions respectively). The group MAVD values increased as impact forces increased from 151.04 deg•s^{-1} for the 10% slower speed condition to 185.94 deg•s^{-1} for the 20% faster speed condition, and from 166.58 deg•s^{-1} for the self-selected pace (no obstacle) condition to 187.53 deg•s^{-1} for the 15% obstacle condition.

Paragraph Number 30 A careful examination of the rearfoot and the knee angle mean ensemble curves revealed that some rearfoot curves had two distinct minimums and a well defined maximum in between (bimodal curves), while others exhibited a single minimum value (unimodal curves). For all subjects, bimodal curves were more obvious for the conditions with greater GRIF values. This can be seen in Figure 2 where rearfoot and knee angle mean ensemble curves are shown for the obstacle conditions of a representative subject (S8). It can be seen from these data that the knee angle curves are minimally affected by the different obstacles and the corresponding GRIF increases, while the rearfoot curves changed from a unimodal to a bimodal formation. To eliminate any uncertainty that the bimodal phenomenon may have been a result of the averaging process to generate the mean ensemble curves, and to further investigate the reasons for the formation of the bimodal rearfoot curves, the two components of the rearfoot angle are presented in Figure 3. As previously mentioned, the rearfoot angle is the difference between the foot angle and the leg angle. The foot angle is the angle formed by the foot and the horizontal, while the leg angle is the angle formed by the shank (lower leg) and the horizontal. As it can be seen from Figure 3, the first peak of the rearfoot angle curve is due to the leg angle, while the parabolic configuration is due to the foot angle. Furthermore, the relationship between GRIF increases and the leg angle are shown in figure 4. It can be seen that the greater the impact forces, the more evident is the first peak on the leg angle curve and consequently on the rearfoot angle curve. These patterns were present in all subjects.

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Paragraph Number 31 Subject means and standard deviations across conditions and across test sessions for curve correlations (CC), and the mean absolute velocity differences (MAVD) are presented in Table 2. As previously described, these biomechanical results were used to construct two corresponding subjects' rankings for susceptibility to injury. The biomechanical rankings along with the clinical evaluation ranking are also presented in Table 2. The rank order comparisons resulted in a significant correlation (r_s = 0.798 with 64% explained variance) between the clinical rankings and the VD rankings. No significant correlation was found between the clinical and the CC (r_s = 0.583 with 34% explained variance) rankings.

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DISCUSSION

Paragraph Number 32 The results indicated that GRIF increased with increases in obstacle heights and speeds (Table 1). However, as shown by the group analysis, these increases in obstacle heights and speeds did not affect TMKF. Lack of changes for TMKF between conditions was also evident in the inspection of the knee angle mean ensemble curves. Curves from a representative subject clearly demonstrated that the knee angle curves were minimally affected by the different obstacles and the corresponding GRIF increases (Figure 2). Group analysis revealed significant changes for TMP but only for the obstacle conditions, and no significant differences were found overall for |TMP-TMKF|. However, the lack of statistical differences for TMP and |TMP-TMKF| may have been caused by the generation of a second minimum on the rearfoot angle curve (Figure 2). The presence of bimodal rearfoot angle curves resulted in dilemma during the identification (selection) of the TMP especially when the two minimums were similar in value. The problematic selection of the TMP due to the bimodal rearfoot angle curves resulted in large TMP standard deviations (Table 1). In addition, standard deviations for |TMP-TMKF| were almost doubled from the self-selected condition to the conditions with the highest GRIF (10% obstacle and 15% obstacle; Table 1). Large TMP standard deviations were also reported by Clarke et al. (5) but an explanation was not presented.

Paragraph Number 33 The increases in obstacle heights and speeds also produced greater MAVD group mean values and augmented the differences between rearfoot and knee angular velocities (Table 1). Large differences between the two velocities indicate antagonistic relationships at the two ends of the tibia and possibly injurious situations. The curve correlations were also affected by the increases in obstacle heights and speeds and the corresponding GRIF increases (Table 1). More specifically, the lowest curve correlations were revealed at the two conditions with the highest GRIF values, the 10% and the 15% obstacle conditions. Decreases in curve correlation values were due to the changes on the rearfoot angle curves since the knee angle curves remained unchanged (Figure 2). Functionally, the lower the curve correlations, the greater the dissimilarities in timing between the rearfoot and knee angle curves. The curve correlation for the self-selected pace condition was not the greatest as might be expected due to the absence of a perturbation. A possible explanation is that the subjects slightly underestimated their comfortable self-selected pace due to the upcoming unknown experimental experience as a caution of any potential fatigue effects and/or injury. Furthermore, the 10% slower speed condition that exhibited the lowest group GRIF values also featured decreased curve correlations (Table 1). During running, the relationship between energetic requirements and speed is a parabolic function. Thus, the 10% slower speed condition could have imposed increased energy requirements on the subjects that might have resulted in discoordination between joint actions. Therefore, it can be theorized that the selection of a comfortable speed by the subject is based not only on energetic requirements but may also be a function of some criteria for avoiding injuries by optimizing coordination between joint actions.

Paragraph Number 34 The above results supported in general the hypothesis that with increases in GRIF, the actions of subtalar pronation/supination and knee flexion/extension become more asynchronous and at some point the rearfoot curve will change from a unimodal to a bimodal formation. This event is clearly demonstrated in Figures 2, 3, and 4. The occurrence of a bimodal rearfoot angle curve is probably a mechanical phenomenon. The first peak of the curve was due to the leg angle, while the parabolic configuration was due to the foot angle (Figure 3). The leg angle shows the motion of the tibia relative to the ground, while the foot angle shows the motion of the foot relative to the ground. Thus, when the foot comes in contact with the ground in early stance, the tibia deviated laterally and then medially in a quick and sudden fashion around the fixed foot. This phenomenon is propably due to impact as illustrated in Figure 4. The greater the impact, the more evident is the first peak on the leg angle curve (Figure 4), and consequently on the rearfoot angle curve (Figure 2). This phenomenon may be explained as a rebound effect of the tibia due to contact with the ground. The harder is the contact with the ground, the larger is this rebound effect. Conversely to the changes on the rearfoot angle, the knee angle remains temporally unaffected by the increases in impact forces. As a result, mistiming or lack of coordination is introduced between the actions of subtalar pronation/supination and knee flexion/extension.

Paragraph Number 35 Examination of the individual subject rearfoot angle curves also revealed that the change from a unimodal to a bimodal rearfoot angle curve did not occur at the same condition for all subjects. The point of change was quite variable which can be explained by considering individual differential responses. Similar to observed patient differences in the examination room, subjects enter the experimental setup with different physiological and anatomical characteristics as well as experiences which influence their performances. If we make the assumption that the change in the rearfoot angle curve can be an injury mechanism, then the system may try to avoid such a change. For example, an experienced runner may remain more coordinated and change later as impact forces increase, compared to a novice runner. This could be due to a better optimization of the available components (degrees of freedom) involved in the shock absorbing system.

Paragraph Number 36 It is likely that the absence of bimodal rearfoot angle curves in the literature and the lack of interest in the changing formation of the rearfoot angle curve, is the result of oversmoothing. By oversmoothing, a bimodal curve can easily be replaced by a non-representative unimodal rearfoot angle curve. Hamill et al. (10) recommended cut-off frequencies between 15 and 18 Hz when rearfoot data are analyzed. They reported that a cut-off frequency below 15 Hz will attenuate the maxima of the rearfoot data and could cause the experimenter to draw incorrect conclusions (10). In the present study, the cut-off frequencies used were 16-20 Hz for the rear view coordinates. In figure 5, a single trial from the 5% obstacle condition is presented with a range of cut-off frequencies (6 to 18 Hz). It can be seen that for frequencies below 15 Hz, the high frequency (impact phenomena) components of the signal are severely attenuated.

Another reason for the lack of interest in the literature for the changing

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INSERT FIGURE 5 ABOUT HERE

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configuration of the rearfoot angle curve, might be that the evaluation of rearfoot motion has focused on maximum values (position or velocity) and the relationships between maximum values (5, 7, 18, 20). However, the actions of knee flexion/extension and the subtalar pronation/supination are continuous phenomena and using a discrete point analysis might result in a limited evaluation. As van Woensel and Cavanagh (27) suggested, the time to maximum pronation is an unreliable variable and they recommended that a two-phase profile using rearfoot velocity and position is more useful in describing rearfoot motion.

Paragraph Number 37 A possible limitation of this study concerning the rearfoot angle curve is how representative it is since it will be the outcome of a two dimensional biomechanical evaluation, and pronation (rearfoot motion) is a three dimensional phenomena. However, as reported in the literature (1, 10) the differences between the two types of analyses are minimal from foot contact and until completion of 80% of the support period. Differences are increased as the foot moves out of the plane with maximum differences occurring during toeoff which affect the end of the rearfoot angle curve (from 80% to the end of the support phase; 1, 10). Since the key phenomena regarding the rearfoot, foot, and leg angles occur between 15% and 65% of the support period, a two dimensional analysis should be adequate. The results of this study regarding rear view parameters are also similar in value to those reported by others (2, 3, 5, 7, 9, 20, 27).

Paragraph Number 38 A major problem in the prevention of running injuries is the absence of clinical evaluation to validate biomechanical results. As Messier and Pittala mentioned (18), biomechanical studies are usually just descriptive and as a result they fail to establish external validity. In the present study and in an effort to increase the external validity, subjects' rankings for susceptibility were constructed based on an clinical evaluation and the biomechanical results. Stanley L. James MD FACSM, conducted the clinical examination and also ranked the subjects on susceptibility to injury. The clinical tests used are common practice for clinicians and they are well described in the literature (13, 16, 28). However, the subjects' ranking was based on the clinical experience of Dr. James. This method is not based on any previous research and it is subjective. However, if the evaluation of patients in the clinics is currently based on the same criteria (self-perception of symptoms' importance), then it seems logical to attempt to validate the biomechanical results with the clinical evaluation of an experienced orthopedist.

Paragraph Number 39 From the two biomechanical techniques that were used to construct subjects' rankings for susceptibility to injury, only velocity differences produced a significant relationship with the clinical subjects' rankings. No comparison was attempted between the TMP and TMKF differences and the clinical results, due to the discrete point evaluation nature of TMP and TMKF. As previously mentioned, the actions of the two joints are continuous and dynamic. Thus, it is unlikely that the evaluation of only two points (MP and MKF) from the entire support period curve will be representative of these actions. Especially, when the rearfoot angle curve is bimodal, the selection of MP and subsequently of TMP is a very difficult and ambiguous process. The techniques that were based on the analysis of the entire support period (CC and VD) may be more appropriate when continuous phenomena are investigated. It has been previously suggested that rearfoot velocity might be associated with injury mechanisms but limited efforts have been made to relate this variable to knee joint function (5, 7, 20). The results of this study revealed that the velocity differences that provide a reasonable estimate of the antagonistic relationships between the actions of the subtalar and the knee produced the highest correlation with the clinical evaluation (r_s = 0.798). As it has been mentioned previously, fewer than 20% of treatments used in sports medicine are supported by solid scientific evidence (22), and diagnosis, treatment, and prevention of running injuries is reported to be more of a speculative art than a science (14, 17, 22, 26). The overuse of speculation has been attributed to the limited exchange of information between biomechanics and clinical medicine (14, 17, 22, 26). Therefore, the results of this study and especially the parameter of velocity differences as a biomechanical tool may be able to bridge this gap of information exchange and assist in the evaluation of running injuries.

Paragraph Number 40 However, the results of this study did not predict relationships of the biomechanical techniques used and specific running injuries e.g., plantar fasciitis, iliotibial band syndrome. In this study, running injuries were treated as a set of different injuries and it was proposed that the lack of coordination between subtalar pronation/supination and knee flexion/extension is a possible injury mechanism that may lead to many different running injuries. This mechanism may have the potential to predict runners who are susceptible to injury. It is also important that the biomechanical techniques presented in this study and especially the velocity differences will be further investigated and validated via prospective observational studies.

Paragraph Number 41 In conclusion, the results of the study suggested that a possible mechanism responsible for various injuries to runners may be lack of coordination between the actions of subtalar pronation/supination and knee flexion/extension. Lack of coordination may result in an antagonistic relationship between these actions via tibial rotation. Furthermore, the results of the subtalar-knee relationship model reveal that differences can be augmented with increases in impact force which over time could produce running injuries. In addition, this mechanism showed some potential for providing similar results for susceptibility to injury assessed through clinical evaluation. Particularly, the velocity differences between the actions of the subtalar joint and the knee, which in essence capture the antagonistic nature of their relationship, produced the best comparative results.

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Figures Legends

Figure 1. Sagittal, rear view marker locations and angle identification.

Figure 2. Rearfoot and knee angle curves from subject 8 for the obstacle conditions. Each curve is an ensemble average over all trials. The self-selected pace without an obstacle condition is represented with a solid line, while the obstacle conditions with perforated lines (5% alternate dashes and dots; 10% dashes; 15% dots) . The top group of curves are the knee curves, and the bottom group of curves are the rearfoot curves.

Figure 3. Foot and leg angle curves from subject 8 for the obstacle conditions. Each curve is an ensemble average over all trials. The self-selected pace without an obstacle condition is represented with a solid line, while the obstacle conditions with perforated lines (5% alternate dashes and dots; 10% dashes; 15% dots). Both angle curves are plotted with the same axis, since they have comparable values. The top group of curves are the foot curves, and the bottom group of curves are the leg curves.

Figure 4. Leg angle and ground reaction forces curves from subject 8 for the obstacle conditions. Each curve is an ensemble average over all trials. The self-selected pace without an obstacle condition is represented with a solid line, while the obstacle conditions with perforated lines (5% alternate dashes and dots; 10% dashes; 15% dots). The top group of curves are the force curves, and the bottom group of curves are the leg curves.

Figure 5. Rearfoot angle curves for the same trial and from the 5% obstacle condition smoothed with different cut-off frequencies.

APPENDIX A

The form that was used by Stanley L. James (MD, FASCM) to conduct the clinical examination. R stands for right side and L for left side. In addition, information was gathered regarding the subject’s running experience and injury history.

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1. Angle of Gait
a. In-toe R L
b. Straight R L
c. Out-Toe R L

2. Genu Varum
Valgum
Straight

3. Tibial Torsion
a. None
b. Internal
c. External
d. Squinting Patellae

4. Leg Varus (to floor) R L

5. Foot Type Standing
a. High Arch R L
b. Low Arch R L
c. Medium R L

6. Standing Pronation
a. Maximum R L
b. Add. Avail R L

7. Extremity Length R L

8. Ankle Dorsiflexion
a. Knee Extended R L
b. Knee Flexed R L

9. Hip Rotation (Prone)
a. Internal R L

10. Foot Motions
a. Loose R L
b. Tight R L
c. Normal R L

11. Subtalar Joint
a. Loose R L Inversion R L
b. Normal R L Eversion R L
c. Restricted R L
d. Varus R L
e. Neutral R L
f. Valgus R L

12. Forefoot Alignment
a. Neutral R L
b. Varus R L
c. Valgus R L

13. Position 1st Ray
a. Plantarflexed R L
b. Dorsiflexed R L
c. Neutral R L

14. Motion 1st Ray
a. Normal R L
b. Mod. Restr. R L
c. Restricted R L

15. 1st MPJ Motion
a. Restricted R L
b. DorsiflexionR L
c. Painful R L
d. Tender R L

16. Toe Position

b. External	R	L	a. Straight	R	L	
c. Symmetrical	R	L	b. Contracted	R	L	
			c. Subluxed	R	L	
			d. Hall. Val.	R	L	17. Location Callus/Corns