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Subtalar and knee joint interaction during running at various stride lengths

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ABSTRACT

Background: It has been suggested that during running proper coordination between subtalar pronation/supination and knee flexion/extension via tibial rotation is important to attenuate ground reaction impact forces (GRIF). Lack of coordination over time may produce a wide range of injuries. It was hypothesized that increasing stride length would result in higher GRIF. It was also hypothesized that alterations in stride length would result in changes of the subtalar/knee coordination.

Methods: Six subjects ran under three different stride lengths (normal stride, understride and overstride) at their self-selected pace. Sagittal, rear view kinematic data and GRIF kinetic data were collected. The subtalar/knee coordination was evaluated via timing and relative velocity measures. Repeated measures ANOVA were performed on these measures with a Tukey post-hoc analysis conducted where appropriate (p<0.01).

Results: Increased stride length produced significant increases in GRIF and significantly augmented the differences between rearfoot and knee angular velocities. A change in the rearfoot angle curve from a unimodal (one minimum) to a bimodal (two minimums) parabolic configuration was also observed. The appearance of the additional minimum was attributed to the increased impact with the ground.

Conclusions: The results indicated that increases in GRIF via changes in stride length could disrupt the coordination between subtalar and knee joint actions.

Keywords: Coordination, Running Injuries, Rearfoot, Pronation, Stride Length
INTRODUCTION

One of the most popular forms of physical activity is running. Unfortunately, running often causes injuries to the lower extremities. The yearly incidence of injuries among runners is estimated to be between 37 and 56%.\textsuperscript{1} About 70-80% of these injuries are of an overuse type and involve the knee, leg, ankle, and foot.\textsuperscript{1-2} To prevent and treat such injuries it is essential to understand the mechanisms that predispose and lead to injury.

However, the medical literature suggests that there has been little progress in our understanding of running injury mechanisms, and the exchange of information between biomechanics and clinical medicine has been limited.\textsuperscript{1, 3-4} In an extensive review of the running injuries literature, van Mechelen\textsuperscript{1} identified four factors significantly related to running injuries: a) injury proneness, b) lack of experience, c) running to compete, and d) excessive weekly running distance. Furthermore, it has been suggested that a dynamic, functional abnormality may be more important than a static malalignment in predisposing a runner to injury.\textsuperscript{1, 5}

Increased impact forces have also been associated with running injuries.\textsuperscript{1, 3, 6} However, it remains unclear how impact forces can actually cause running injuries.\textsuperscript{1, 6} Two natural mechanisms available to attenuate these forces are knee flexion and subtalar pronation. Specifically, the knee joint has been credited for producing a 70% reduction of the impact peak amplitude.\textsuperscript{7} Subtalar pronation allows impact forces to be attenuated by increasing the time during which the foot becomes stationary on the ground.\textsuperscript{8-10} Bates et al.\textsuperscript{11} suggested that decreased subtalar pronation could cause impact forces to be abruptly
attenuated by the supporting structures possibly resulting in injury.

Less attention has been directed toward the coordinative action between the subtalar and knee joints. As described by Bates and colleagues,\textsuperscript{11-12} pronation usually occurs in the first 50\% of stance. During pronation the knee joint flexes. The synchronous actions of pronation and knee flexion during the first part of stance allow for much of the impact forces to be attenuated. Following midstance, the foot supinates and the knee extends through take-off. Pronation and knee flexion during early stance are accompanied by an internal rotation of the tibia, while supination and knee extension during late stance are associated with external tibial rotation.\textsuperscript{9-10, 13} Asynchronous timing or lack of coordination between the subtalar and knee joint movements can result in an antagonistic relationship between these joints via tibial rotation. Such a relationship could result in excessive stress on soft tissues and, after multiple repetitions as is the case in running, could result in injury.

As Edington et al.\textsuperscript{14} reported, there is limited information comparing the rearfoot motion to other timing events. They suggested that this topic deserves much more attention. Recently, Payne\textsuperscript{15} indicated that the pathologic mechanisms behind the known relationship between excessive subtalar joint pronation and patellofemoral dysfunction have not yet been elucidated. Hamill et al.\textsuperscript{16} reported timing differences between subtalar and knee joint actions due to decreased shoe hardness. Lately, Stergiou and Bates\textsuperscript{17} and Stergiou et al.\textsuperscript{18} observed that increases in impact forces due to increased surface hardness, speed and obstacle clearance augmented timing discrepancies between subtalar and knee joint function. The increased impact forces also resulted in a transition of the rearfoot angle
curve from a unimodal to a bimodal configuration. 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 maximum in between. The authors suggested that when a bimodal rearfoot angle curve is coupled with the typical unimodal knee angle curve, lack of coordination between the subtalar and knee joint actions is more likely to occur. The above authors also suggested that increased impact forces may facilitate abnormal timing and disrupt the coordinative actions of the subtalar and knee joints which eventually can lead to running injuries.

Variations in stride length occur frequently during running. Thus, stride length and its relationship with stride frequency and speed have been investigated extensively in the running literature. Changes in stride length have also been reported to influence impact forces. However, no studies have examined the effect of stride length on subtalar pronation. In addition, it is unclear if changes in stride length can elicit abnormal timing and disrupt the coordinative actions of the subtalar and knee joints like the other perturbations mentioned above.

Thus, the purpose of the present study was to examine the effect of stride length alterations on the relationship between subtalar and knee joint actions. It was hypothesized that increasing stride length would result in higher ground reaction forces. It was also hypothesized that this perturbation would result in changes of the subtalar/knee coordination as measured by timing and relative velocity measures.
MATERIALS AND METHODS

Six healthy male recreational runners (mean(s.d.) age 22.2(3.5) years; height 176(6.4) cm; mass 70.3(7.1) kg) running a minimum of 10 miles per week for at least 1 year served as subjects. All subjects exhibited a heelstrike pattern during running at a self-selected pace. Prior to testing, each subject read and signed an informed consent document approved by the University of Oregon Human Subjects Review Board.

A force platform (model OR6-5-1; AMTI, Watertown, Massachusetts, USA) was used to collect the ground reaction force (GRF) data. The force platform was installed in the middle of a 40 m runway in the Biomechanics Laboratory at the University of Oregon. A signal conditioner/amplifier (model SGA6-3; AMTI) was employed in conjunction with the force platform. The signal conditioner/amplifier was interfaced with a sampling system (APAS; ARIEL Dynamics, Trabuco Canyon, California, USA) interfaced to a computer. One force channel (Fz; the vertical component) and one synchronizing channel, were collected at a sampling rate of 1000 Hz.

Kinematic data were collected (200 Hz) using two NEC (NEC America, Woo Dale, Illinois, USA) high speed video cameras interfaced to a real time automated video based tracking system (Motion Analysis, Santa Rosa, California, USA). The cameras were positioned to obtain a side (sagittal) and rear (frontal) view of the right lower extremity during the stance period. Camera distances were 14 and 11 meters, respectively, each equipped with a zoom 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. The proper positioning of these markers was described in detail elsewhere.\textsuperscript{14,17} The reflective markers were illuminated with 650W lamps located directly above the camera lens. The retroreflective images from each camera were obtained and translated to cartesian coordinates using a video processor (VP320; Motion Analysis) interfaced to a computer. Data collection by the APAS and the video tracking system was triggered by a manual transistor/transistor/logic (TTL) switch to synchronize the video and force data.

All subjects were asked to perform a stride perturbation protocol. Running speed was monitored over a 3 m interval using a photoelectronic timing system (model 63520; Lafayette Instr., Lafayette, Indiana, USA). 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 the force platform. During warm-up the subject established a comfortable running pace which was recorded. This speed ($\forall 5\%$) was used as a baseline speed for subsequent testing. The average speed from all subjects was 4.58 m/s. 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.

All subjects were asked to run at their previously established baseline pace under three different conditions. The first condition was simply unperturbed running (normal stride; NS). For the other two conditions the subjects were asked to contact the force platform with a shortened stride (understride; US) and an elongated stride (overstride; OS).
This procedure was accomplished as follows. After each subject established his baseline pace and while running, a right foot marker was placed one stride length before the force platform. The marker was moved the length of the subject’s foot closer and farther from the force platform for the US and OS conditions, respectively. The length of the subject’s foot was measured along the anterior-posterior axis for each subject and the mean (s.d.) was determined to be 0.30(0.03) m. This distance was established via pilot work which indicated that to maintain a heelstrike pattern, the US/OS conditions could not deviate from the NS condition by more than one foot length. For both the US and the OS conditions, subjects were instructed to target both the marker and the force platform. Each condition consisted of twenty trials and the order of presentation of the conditions was randomized. Fatigue effects were minimized by the 1 min inter-trial rest period allowed.

The first peak value (impact force peak) of the vertical GRF (Fz) was identified for each trial via laboratory software. Using this software, the user inspected each curve and through an interactive algorithm identified the maximum values. Subsequently, the vertical ground reaction impact force (GRIF) peak values were normalized for body weight, and mean values were calculated across trials for each subject-condition along with the group means for each condition.

The kinematic coordinates were smoothed using a Butterworth low-pass filter with a selective cut-off algorithm based on Jackson. 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 coordinates the following angles were calculated (Figure 1): a) sagittal knee
angle, and b) frontal angles of the foot and the leg with respect to the left horizontal. The rearfoot angle was calculated by subtracting the leg angle from the foot angle. The angular position data were differentiated using a cubic spline routine to calculate angular velocities. The vertical GRF(Fz) and all kinematic parameter data files were normalized to 100 points for the stance period using a cubic spline routine to enable mean ensemble curves to be derived for each subject-condition.

**INSERT FIGURE 1 ABOUT HERE**

The times of occurrence of the minimum knee angle and the maximum negative rearfoot angle 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). Subsequently, the absolute differences between TMP and TMKF (|TMP-TMKF|) were identified for each trial. The mean values of |TMP-TMKF| were calculated for each subject-condition. Group mean values for all subjects were also calculated for each condition. This parameter was used to evaluate timing differences between the actions of the two joints.

An additional parameter was calculated by subtracting the corresponding data points of the rearfoot and knee angular velocity mean ensemble curves: $\omega_{\text{DIFF}} = \omega_{\text{KNEE}} - \omega_{\text{REARFOOT}}$. Using this technique, a new curve was generated that represented the angular
velocity differences throughout the stance period. The maximum negative peak value of the velocity difference curve was identified for each trial via the same interactive process used for the GRIF value. The maximum negative peak value was selected based upon pilot work and related literature\textsuperscript{17} indicating that this point was the most consistent and distinct peak on the angular velocity difference curves. Furthermore, this peak occurs almost immediately after the impact with the ground and approximately at 20\% of stance.\textsuperscript{17} Thus, its occurrence is closely related to the occurrence of the GRIF parameter. The mean values of the maximum velocity differences (MVD) values were calculated for each subject-condition. Group means of all subjects were also calculated for each condition.

Lastly, to examine the actions of subtalar pronation/supination and knee flexion/extension over the entire stance curve correlations were calculated. The curve correlation technique was introduced by Derrick et al.\textsuperscript{24} 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, while a lower correlation revealed the opposite. Mean values for all subjects and each condition were also calculated from these data.

One-way repeated measures ANOVAs (stride by subjects) were performed using the mean values for GRIF, [TMP-TMKF], MVD and curve correlations (CC). In tests that resulted in a significant F-ratio (p<0.01), a Tukey multiple comparison test was performed to identify the location of the significant differences.
RESULTS

The GRIF group results were statistically different \(( F(2,5) = 10.83, p = 0.003, \text{Stat. Power} = 0.95)\), with the post-hoc analysis revealing statistical differences between OS and the other two conditions (Table 1). In addition, it can be observed that the longer the stride length, the greater the impact force (US: 1.677 BW; NS: 1.737 BW; OS: 2.185 BW).

**INSERT TABLE 1 ABOUT HERE**

The MVD group results were also statistically different \(( F(2,5) = 7.61, p = 0.009, \text{Stat. Power} = 0.80)\), with the post-hoc analysis revealing statistical differences between the OS and the other two conditions (Table 1). The NS condition had the lesser MVD group mean value compared to the other two conditions. No statistical differences were found among the curve correlations \(( F(2,5) = 1.173, p = 0.358, \text{Stat. Power} = 0.10)\) and the \(|\text{TMP-TMKF}|\) group results \(( F(2,5) = 1.219, p = 0.335, \text{Stat. Power} = 0.10; \text{Table 1})\).

However, it can be noted that the lesser \(|\text{TMP-TMKF}|\) group mean and standard deviation values were present for the NS condition. The NS condition also exhibited the greater CC group mean value.

Rearfoot and knee angle mean ensemble curves are displayed for all conditions for two representative subjects (S1 in Figure 2; S6 in Figure 3). An examination of these 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 both subjects, bimodal curves can be seen for the OS condition. It is also evident that the configuration of the knee angle curves is less affected (remained
unimodal) by the changes in stride length, while the rearfoot curves changed from unimodal to bimodal for the OS condition. The figures also revealed that the perturbation did not affect all subjects to the same degree. For S1 the increase in impact forces was greater and was reflected more clearly on the rearfoot curves. While all subjects exhibited the change from a unimodal to a bimodal curve during the OS condition, this transition was not as dramatic as for S1. For S6 who represents the other extreme, change occurred but it was not as drastic.

**INSERT FIGURES 2 AND 3 ABOUT HERE**

To further investigate the bimodal phenomena, the rearfoot angle and the vertical GRF (Fz) mean ensemble curves are presented (Figures 4 and 5) for the same subjects (S1 and S6). These data indicated that an increase in impact force (first peak of the Fz curves) was associated with an increase in rearfoot angle (Figures 4 and 5). In addition, a comparison between Figures 4 and 5 demonstrates that the larger the increase in impact force, the larger the effect on the rearfoot angle curves. Furthermore, the increase in impact force was also reflected on the velocity difference curves (Figure 6). In fact, the greater the impact force, the greater the MVD value (negative peak). These patterns were present for all subjects.

**INSERT FIGURES 4, 5 AND 6 ABOUT HERE**
DISCUSSION AND CONCLUSIONS

The results indicated that increases in stride length resulted in greater impact forces. This result supported our first hypothesis. It is also in agreement with the literature which indicated that longer stride lengths are associated with greater impacts.\textsuperscript{22}

The increased impact forces could have resulted in the significant increases observed for the MVD values. However, someone can argue that the MVD changes may not be the result of the increased impact forces, but they are related to changes in the kinematics due to the increased stride length. Although this is a valid argument, it is probably not the case. As it is indicated in Figure 6, the MVD values occurred almost immediately after the impact peaks and approximately at 20\% of stance. In addition, these velocity differences curves are similar to curves presented by Stergiou and Bates,\textsuperscript{17} where the MVD values occurred at similar times. Thus, the changes in the MVD values were probably associated with the increased impact forces.

It has been previously suggested that rearfoot velocity might be associated with injury mechanisms,\textsuperscript{8, 14} but limited efforts have been made to relate this variable to knee joint function. Recently, Stergiou et al.\textsuperscript{18} found that velocity differences between the actions of the subtalar and knee joint correlated significantly with a clinical evaluation regarding susceptibility to injury. Functionally, large MVD values indicate increased differences between the knee and rearfoot angular velocities. Increased velocity differences may also suggest antagonistic relationships at the opposite ends of the tibia. It should be noted that the knee and rearfoot motions are in orthogonal planes (sagittal and frontal) to
each other. However, both motions result in tibial rotation in the transverse plane.\textsuperscript{9-10,13} The subtalar joint, functioning as a mitered hinge connection, transmits the rotation of the foot in the longitudinal axis, to rotation of the tibia about its long axis. At the tibiofemoral joint, the asymmetric condyles cause the tibia to rotate around the femur during flexion and extension. Therefore, it is logical to study the relationships at the two ends of the tibia via differences in knee and rearfoot velocities.

No statistical differences were found for |TMP-TMKF|. However, the lack of statistical differences could have resulted from the generation of the second minimum value on the rearfoot angle curve (Figure 3). The presence of bimodal curves results in ambiguity between the events being evaluated especially when the two minimums are similar in value. Therefore, the problematic selection of the TMP resulted in large |TMP-TMKF| standard deviations (Table 1). Specifically, both US and OS conditions had much larger standard deviations than the NS condition. Large TMP standard deviations have also been reported in the literature\textsuperscript{8} but no explanations are given.

No significant differences were found for the CC values (Table 1). However, it can be observed that the greatest correlation was at the NS condition indicating similar shapes for the two curves, and thus, a more synchronous relationship between the actions of the subtalar and the knee joints. Furthermore, the NS condition produced the greatest CC, the smallest MVD, and the smallest |TMP-TMKF|. Greater correlations, lesser velocity differences, and smaller timing differences can all be associated with a more synchronous relationship between the two joints or in other words with better coordination. Therefore, it
can be theorized that the selection of a comfortable pace with a specific normal stride could be a function of some criteria for optimizing coordination between joint actions. Such a theory can be explored in future studies.

The increases in impact force resulted in a unimodal to bimodal change in the rearfoot curves (Figures 4 and 5). This observation is consistent with the results from other studies, where the rearfoot curve was similarly affected by increased impact forces due to changes in surface hardness, speed and obstacle clearance. The increased impact with the ground can cause rapid pronation during early stance, disturbing the natural and synchronous actions of the subtalar and knee joints. This early rapid pronation has to quickly be followed by supination. However, the knee is still flexing. Supination and flexion are opposite movements regarding tibial rotation. Therefore, over time and with multiple repetitions as it is the case in running, these opposite movements can cause stresses to the lower extremity.

The different responses among subjects (S1 versus S6) are due to individual variability. Subjects exhibit different physiological and anatomical characteristics that can influence their performances. Furthermore, if the change in rearfoot angle is related to an injury mechanism, then the system might try to avoid such a change. A better runner can possibly optimize more efficiently the available shock absorbers to resist this possible change.

The lack of interest in the literature for the changing formation of the rearfoot angle curve could be due to oversmoothing. An oversmoothed bimodal curve can be mistaken for
a unimodal curve.\textsuperscript{18} The literature\textsuperscript{18, 25} suggests that for rearfoot data the cut-off frequencies used should be between 15 and 18 Hz, while lower cut-offs will lessen maxima and will cause false conclusions by the experimenter. Therefore, extreme care must be given to the selection of the cut-off frequencies to avoid distorting the data. In the present study, the cut-off frequencies used were between 16 and 20 Hz for the rear view coordinates.

A possible limitation of this study regarding the rearfoot data is the use of a two-dimensional analysis to describe the three-dimensional rearfoot motion. However, the literature\textsuperscript{25-27} reports that the differences between the two types of analysis are minimal after foot contact through 80\% of stance. Since the important phenomena of the changing rearfoot angle occur between 15 and 65\% of stance, the authors believe that a two-dimensional analysis is adequate. However, future studies should consider validating our results with a three-dimensional analysis.

In this study, it was proposed that a possible injury mechanism which can lead to a number of different running injuries, may be the lack of coordination between the actions of subtalar pronation/supination and knee flexion/extension. This lack of coordination can be augmented by increases in impact forces. Changes in stride length can elicit such increases. However, it is important to further investigate and validate the results of the study via prospective observational studies.
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Table 1. Group means and SDs for all conditions and for all variables evaluated.

<table>
<thead>
<tr>
<th>Condition</th>
<th>GRIF (BW)</th>
<th>[TMP-TMKF] (% of support)</th>
<th>MVD (deg/s)</th>
<th>CC (Pearson r)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Under Stride</td>
<td>1.677 OS + 0.148</td>
<td>11.23 ± 7.13</td>
<td>-587.25 OS + 125.92</td>
<td>0.57 ± 0.29</td>
</tr>
<tr>
<td>Normal Stride</td>
<td>1.737 OS + 0.320</td>
<td>6.50 ± 1.63</td>
<td>-569.81 OS + 152.15</td>
<td>0.71 ± 0.29</td>
</tr>
<tr>
<td>Over Stride</td>
<td>2.185 OS + 0.258</td>
<td>10.53 ± 5.69</td>
<td>-705.54 OS + 187.15</td>
<td>0.61 ± 0.28</td>
</tr>
</tbody>
</table>

GRIF = ground reaction impact force, [TMP-TMKF] = absolute differences between time to maximum pronation and time to maximum knee flexion, MVD = maximum velocity differences, CC = curve correlations.

Condition means that are significantly different (p<0.01) are shown in superscripts.
Figure Legends

**Figure 1.** Sagittal, rear view angle identification.

**Figure 2.** Rearfoot and knee angle curves from subject 1 for all conditions. Each curve is an ensemble average over all trials. The NS condition is represented with a solid line, while the other two conditions with markers on lines (US circles and OS crosses). The top group of curves are the knee curves, and the bottom group of curves are the rearfoot curves.

**Figure 3.** Rearfoot and knee angle curves from subject 6 for all conditions. Each curve is an ensemble average over all trials. The NS condition is represented with a solid line, while the other two conditions with markers on lines (US circles and OS crosses). The top group of curves are the knee curves, and the bottom group of curves are the rearfoot curves.

**Figure 4.** Rearfoot angle and Fz curves from subject 1 for all conditions. Each curve is an ensemble average over all trials. The NS condition is represented with a solid line, while the other two conditions with markers on lines (US circles and OS crosses). The top group of curves are the force curves, and the bottom group of curves are the rearfoot curves.
**Figure 5.** Rearfoot angle and Fz curves from subject 6 for all conditions. Each curve is an ensemble average over all trials. The NS condition is represented with a solid line, while the other two conditions with markers on lines (US circles and OS crosses). The top group of curves are the force curves, and the bottom group of curves are the rearfoot curves.

**Figure 6.** Velocity differences and Fz curves from subject 1 for all conditions. Each curve is an ensemble average over all trials. The NS condition is represented with a solid line, while the other two conditions with markers on lines (US circles and OS crosses). The top group of curves are the force curves, and the bottom group of curves are the velocity differences curves.