IS REARFOOT PRONATION A SHOCK ATTENUATING JOINT ACTION?

Michelle Countryman[1, 2], Kristian O'Connor[2], Joseph Hamill[2]
[1]Fila Sport, Peabody, MA, USA
[2]Biomechanics Laboratory, University of Massachusetts, Amherst, MA, USA

The purpose of this study was to investigate the hypothesis that sub-talar joint pronation (i.e. rearfoot motion) is a shock attenuating mechanism during locomotion. Forty-seven males and 48 females served as subjects by walking/running on a treadmill at 6 different locomotor speeds while instrumented with a tibial accelerometer and a rearfoot goniometer. Correlations were performed between rearfoot and impact shock parameters. The results indicated that peak g was negatively correlated with maximum rearfoot angle (r=-0.35) and positively correlated with total rearfoot motion (r=0.60). However, in both cases, the common variance between the parameters was low. It must be concluded that the actions of rearfoot pronation during locomotion are, at best, a peripheral but not a major shock attenuating mechanism.

KEY WORDS: rearfoot motion, pronation, impact shock, accelerometry.

INTRODUCTION: Rearfoot motion has been used to define pronation/supination of the subtalar joint. In this study, rearfoot motion was estimated using the measure of calcaneal eversion/inversion with respect to the leg. The subtalar joint is located between the talus and the calcaneus. During this motion, the talus functions as a part of the leg, whereas during talocrural (ankle) dorsiflexion/plantar flexion the talus functions as a part of the calcaneus. Since the subtalar joint is not aligned with any of the cardinal planes of the body, pronation of the subtalar joint appears to have components of calcaneal eversion with respect to the leg, forefoot abduction and dorsiflexion. Conversely, subtalar supination appears to be made up of calcaneal inversion, forefoot adduction and plantar flexion. Although no research has been done to support or refute the idea, it has been proposed by several researchers that pronation acts as an impact shock attenuator (Knutzen and Price, 1994; Milani et al., 1995). There are multiple forces acting on the foot and leg during the stance phase of gait such as compression, rotation, anterior and medial shear. The most prominent of the forces supposedly affected by subtalar pronation occurs during the impact phase of stance usually within the first 50 ms after foot contact. These vertical loads are termed impact shock, and reach up to 2.32 BW at rates of up to 113 BW/s (Munro et al., 1987). Impact shock characteristics such as high peak magnitudes or temporal abnormalities have also been attributed to complications in runners. Such complications include iron depletion, overuse injuries and joint degeneration (Voloshin and Wosk, 1982; Falsetti et al., 1983; Simon et al., 1972). Therefore, the purpose of this study was to investigate the relationship between rearfoot motion and impact shock. Specifically, it was hypothesized that a greater range of motion in the rearfoot would result in lower impact shock values and thus greater shock attenuation.

METHOD: Subjects for this study were 47 males with a mean height of 1.75 m (+/- 0.05), a mean mass of 71.0 kg (+/- 7.55) and a mean age of 20.8 years (+/- 3.16) and 48 females with a mean height of 1.66 m (+/- 0.07), a mean mass of 59.5 kg (+/-7.58) and a mean age of 20.7 years (+/-3.39). Only subjects who reported no previous lower extremity injury requiring a doctor's care, no pain at the time of testing and read and signed informed consent documents were included. These subjects could be classified as recreational runners. Throughout the testing procedure, subjects walked and ran on a motorized treadmill. An electrogoniometer was used to measure the rearfoot angle (calcaneal inversion/eversion). The electrogoniometer consisted of one fixed arm and one movable arm connected by a potentiometer. The bottom fixed arm of the electrogoniometer was attached adhesively to the posterior heel of the running shoe. The top movable arm of the electrogoniometer was placed within a mechanical guide attached adhesively to the posterior aspect of the leg. This guide allowed the top arm of the goniometer to slide up and down freely but not side-to-side. Thus, the potentiometer measured the angle between the two arms of the goniometer (the estimated calcaneal inversion/eversion angle with respect to the leg). Out-of-plane motion was taken into account because motion in all planes was free to occur but only motion in the frontal plane was measured. The joint at the potentiometer was allowed to flex in dorsi-/plantar flexion and internal/external rotation. A 1.7-gram PCB Piezotronics single axis accelerometer was mounted on the distal anterio-medial aspect of the tibia using adhesive tape and an elastic strap. The axis of the accelerometer was aligned vertically. An additional single-axis accelerometer was mounted under the belt of the treadmill and connected to the microcomputer. This treadmill-mounted accelerometer was used to monitor foot strikes of both feet and served as a timing verification. The electrogoniometer and the two accelerometers were interfaced to a microcomputer via a 12 bit A/D converter and each was sampled at 1000 Hz. Prior to beginning the test protocol, subjects first read and signed informed consent documents. Subjects were then fitted with laboratory shoes of the same model, durometer and midsole density in order to avoid any effects due to differences in footwear. A number of differences in footwear such as midsole durometer, density, varus/valgus slant and lateral heel flare have been shown to affect rearfoot motion parameters (e.g. Nigg & Morlock, 1987; van Woensel & Cavanagh, 1992). Subjects were then fitted with the rearfoot electrogoniometer and an accelerometer on the right lower extremity. Locomotor speeds were presented in a balanced order to all subjects. Walking speeds consisted of 1.03 m/s, 1.52 m/s and 2.01 m/s. Running speeds consisted of 2.68 m/s, 3.58 m/s, and 4.47 m/s for males and 2.68 m/s, 3.35 m/s and 4.03 m/s for females. At each locomotor speed, subjects walked or ran for approximately one minute after which three 5-second data-capture trials including approximately 5-10 foot falls were collected. An acceptable trial included the subject completing the run or walk without complication (including tripping or equipment failure). At the completion of the walking and running bouts, a calibration trial was collected that yielded the standing rearfoot position for each subject. The sampled data from the electrogoniometer were filtered using a fourth order zero-lag low-pass Butterworth filter with a cut-off frequency of 18 Hz. This cut-off frequency was determined using a residual analysis. The resulting data were then converted from volts into degrees. The calibration or stance trial from the electrogoniometer data were subtracted from each trial at each speed for each subject. This counteracted errors due to the estimated placement of the equipment by the investigator. The sampled data from the accelerometers were converted from volts into units of acceleration, or g's (1g = 9.81 m/s^2). The stance phase of at least ten steps at each locomotor speed was extracted. Heel strike was defined as occurring when the acceleration from the treadmill accelerometer reached 20% of its maximum acceleration. Parameters describing rearfoot motion were determined for each stance phase of each condition for each subject. Maximum pronation angle (MP) and total range of rearfoot motion (TRM) were calculated from the electrogoniometer data. In addition, the peak acceleration or peak tibial impact shock (PG) and the time to peak acceleration (TPG) were determined from the accelerometry data. Each of these measures was extracted for each step for each subject and then numerically averaged. A Pearson Product Moment Correlation was conducted to determine whether any relationship existed between the following pairs of variables: PG and TRM, PG and MP.

RESULTS: While there were differences in rearfoot motion between genders at each of the locomotor speeds, for the purpose of this study the data were collapsed across gender. The range of maximum pronation angles ranged from 6.03 to 9.04° for the walking speeds and ranged from -10.13 to -11.70° across the running speeds. Thus, the magnitude of MP tended to increase with increases in speed. Total range of rearfoot motion (TRM) was greater for male subjects than female subjects by approximately 2°. Walking TRM averaged from 8.09 to 13.95° while running TRM ranged from 14.02 to 21.34°. TRM also increased with increases in speed. There was no difference between genders in the acceleration data and so these data were also grouped. Peak g (PG) tended to increase with increases in locomotor speed. PG during walking averaged from 1.14 to 2.16 g and during running from 3.32 to 6.74 g. The correlations conducted in this study were calculated across locomotor speeds. Both correlations were statistically significant ($p < 0.05$) indicating that the correlations were significantly different from zero. MP and PG were negatively correlated (r = -0.35; R^2 = 0.12) and TRM and PG were positively correlated (r = 0.60; R^2 = 0.36). Scatter plots of the two correlations are presented in Figures 1 and 2.

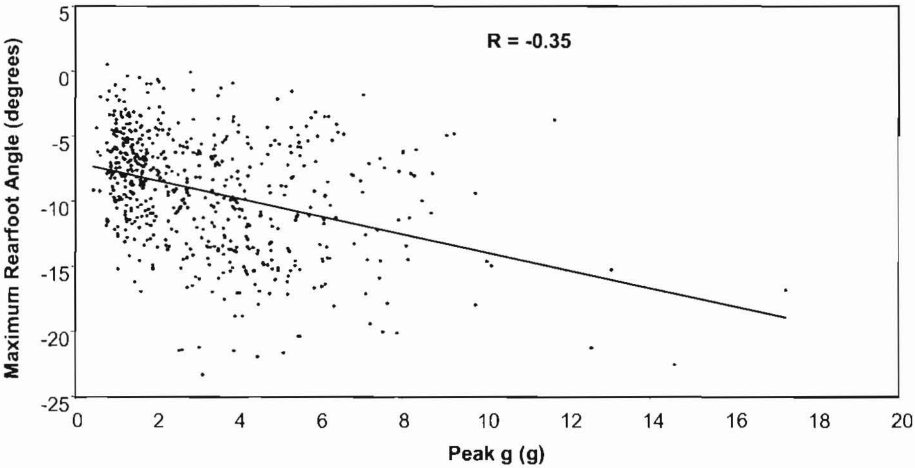

Figure 1. Scatter plot of peak g versus maximum rearfoot angle for all subjects at all locomotor speeds. Greater negative values indicate a greater maximum pronation angle.

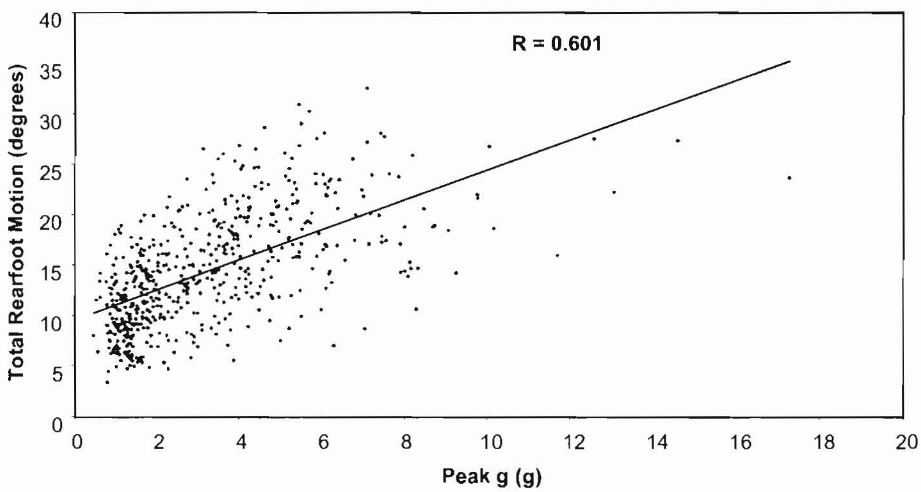

Figure 2. Scatter plot of peak g versus total rearfoot motion angle for all subjects at all locomotor speeds.

DISCUSSION: Several authors have proposed the notion that subtalar pronation is a mechanism that increases shock attenuation (Knutzen and Price, 1994; Milani et al., 1995). It was hypothesized that the act of pronation over a greater range of motion would spread out the time over which ground reaction force was applied to the body. In order to explore this hypothesis, PG was correlated with both MP and TRM using a Pearson Product Moment Correlation. The results indicated that there was little relationship between MP and PG ($r = -0.35$; $R^2 = 0.12$). This is a weak correlation at best with the common variance between the parameters was only 12%. It is also in the opposite direction as to what was hypothesized. It appears, therefore, that the sub-talar joint action from touchdown to mid-stance (i.e. where the maximum rearfoot angle occurs) has little effect on shock attenuation during running. However, there are other joint actions such as the knee (McMahon et al., 1987) that are significant shock attenuation actions. This finding does not appear to support the notion that rearfoot motion is a shock attenuating mechanism. It may be that pronation is merely the consequence of the impact. In the other correlation conducted, TRM increased as PG increased ($r = 0.60$; $R^2 = 0.36$). This seems to indicate that greater amounts of shock occurred as greater ranges of rearfoot motion were achieved. This finding actually contradicts the idea that pronation is a mechanism of shock attenuation. This correlation is also not overwhelmingly strong with a common variance between the parameters of 36%.

CONCLUSION: The results of this study appear to refute the hypothesis that the pronation motion of the rearfoot is a shock attenuating action during locomotion. There are other lower extremity joint actions that are shock attenuating motions such as knee flexion. It is possible, however, that pronation plays only a minor role in shock attenuation.

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