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Effect of Strike Pattern and Orthotic Intervention on Tibial Shock During Running

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The main purpose of this study was to investigate the effects of both strike pattern (forefoot vs. rearfoot strike pattern) and orthotic intervention on shock to the lower extremity. Semi-rigid orthotic devices were manufactured for 15 injury-free recreational runners. Tibial accelerometry, ground reaction force, and 3D kinematic data were collected on their right leg in four conditions: forefoot strike (FFS) and rearfoot strike (RFS) with and without orthotics. Two-way repeated-measures analysis of variance tests were used to assess the effects of strike pattern and orthotic intervention on tibial acceleration; angular excursions of the ankle and knee; ground reaction force (GRF) vertical and anteroposterior peaks and load rates; and ankle, knee, and leg stiffness. There was a significant increase in tibial acceleration for the FFS pattern compared to the RFS pattern. This may be explained in part by the significantly greater peak vertical GRF, peak anteroposterior GRF, anteroposterior GRF load rates, knee stiffness, and leg stiffness found in the FFS pattern compared to the RFS pattern. Tibial acceleration and rearfoot eversion excursions were similar between the orthotic and no-orthotic conditions. Knee flexion excursion and average GRF vertical load rates were significantly decreased while dorsiflexion excursion and knee stiffness were significantly increased in the orthotic condition. No significant interactions were found between strike pattern and orthotic condition for any variables assessed.

Key Words: forefoot, rearfoot, tibial acceleration, stiffness

Introduction

Over the course of a 5-k run, the average runner will strike the ground approximately 3,000 times, placing GRF loads of 2.0 to 3.0 times body weight on each leg

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(Clarke, Frederick, & Cooper, 1983; Hennig & Lafortune, 1991; Hennig, Milani, & Lafortune, 1993; McClay & Manal, 1995b). This repetitive loading of the lower extremity sends shock waves up the leg and has been associated with overuse injuries in runners (Burr, Milgrom, Boyd, et al., 1990; Grimston, Engsburg, Kloiber, & Hanley, 1991). These shock waves are produced from the impulsive loading of GRF resulting from foot strike. Displacement in response to this loading serves to attenuate shock by decelerating the body's center of mass during the loading phase of stance (Clarke et al., 1983). This displacement includes both the strain of tissues across joints and kinematic mechanisms such as flexion at the hip, knee, and ankle.

While some eversion is necessary, excessive eversion is thought to lead to soft tissue injuries of the foot and knee by increasing the strain in these structures (Bates, Osternig, Mason, & James, 1978; Hamill, Bates, & Holt, 1992; Hintermann & Nigg, 1998; McClay & Manal, 1997). Foot orthotic devices (FOD) are used to control excessive motion at the rearfoot, and indirectly the knee, thereby alleviating overuse injuries such as posterior tibial tendonitis and patellar tendonitis. Although limiting eversion through the use of FOD may help reduce these soft tissue injuries, the reduction in movement may result in an increase in tibial shock and subsequently increase the risk of stress-related bony injuries such as tibial stress fractures.

While most runners strike the ground with their rearfoot first, approximately 20% of distance runners make initial contact with the ground at midfoot or forefoot (Kerr, Beauchamp, Fisher, & Neil, 1983). Forefoot strike (FFS) runners have been shown to have increased rearfoot eversion excursions and velocities compared to rearfoot strike (RFS) runners (McClay & Manal, 1995a). Therefore FFS runners may benefit from orthotic intervention. The heel, however, does not make initial contact with the ground in an FFS running pattern. Thus the effect of FOD on rearfoot motion may be negligible when running with an FFS pattern.

In an FFS pattern, the foot lands in plantarflexion and inversion and then dorsiflexes and everts (McClay & Manal, 1995a) using eccentric action of the gastrocnemius, soleus, and tibialis posterior. This eccentric activity, along with the dorsiflexion and eversion excursion, helps decrease the vertical velocity of the center of mass and may effectively attenuate some of the impact shock of foot strike. These mechanisms of FFS running are responsible for the lack, or reduction, of an initial impact peak on the vertical GRF curve that is typically seen in RFS patterns. The absent or reduced impact peak results in lower vertical GRF load rates compared to RFS runners (McClay & Manal, 1995b). Hennig and coworkers (Hennig & Lafortune, 1991; Hennig et al., 1993) reported strong positive correlations between vertical GRF load rates and lower extremity shock, as measured by tibial accelerometers in RFS runners. FFS runners, with their lower vertical GRF load rates, may sustain less tibial shock at foot strike. Therefore RFS runners, with shock related injuries such as stress fractures, might benefit from switching to an FFS pattern.

Oakley and Pratt (1988) compared GRF and tibial accelerometry between RFS and FFS patterns on 18 volunteers running between 3.3 and 3.6 $m \cdot s^{-1}$. Only four of the runners had previous running experience. While Oakley and Pratt did not report peak positive tibial acceleration (PPA), they did find a reduction in peak-to-peak tibial acceleration (PTP) with an FFS pattern compared to an RFS pattern. Unfortunately, the PTP values were graphed in units of millimeters and thus the excursions in gravitational units cannot be deduced.

The primary purpose of this study was to investigate effects of foot-strike pattern (FFS vs. RFS patterns) and orthotic intervention upon tibial acceleration. It was hypothesized that an FFS pattern would exhibit lower tibial acceleration due to the greater joint angular excursions and lower vertical load rates compared to an RFS pattern. It was also hypothesized that the use of foot orthotic devices would increase tibial acceleration in an RFS pattern but not in an FFS pattern, as these devices are hypothesized to reduce the shock-attenuating movement of rearfoot eversion in the RFS pattern only.

Methods

Fifteen RFS runners with no previous use of FOD (average age 22.46 ± 4 yrs; weight 66.41 ± 8.58 kg; height 169.75 ± 6.07 cm) were recruited for this study. Although all were rearfoot strikers, it has been shown that when rearfoot strikers are instructed to run with an FFS pattern, they do not differ in rearfoot kinematics from natural forefoot strikers (Williams, McClay, & Manal, 2000). Therefore, to simplify participant recruitment, rearfoot strikers were used in this study. Prior to their inclusion in the study, all were screened by an experienced physical therapist. Persons with forefoot valgus or varus, limited rearfoot motion (normal range: 10° eversion to 20° inversion), or lower extremity misalignments in standing including genu valgus or varus, excessive rearfoot eversion or inversion, or excessive internal or external rotation of the hip, knee, or ankle were excluded. All runners selected for this study gave an informed written consent in agreement with the university's human subjects review board.

Foot orthotic devices were fabricated from a non-weight-bearing, neutral positioned plaster cast (Losito, 1996). They were constructed from suborthelene and covered with a neoprene pad. As all participants had normal alignment, a common rearfoot varus post of 6° was applied to all devices. No forefoot posting was added since persons with forefoot varus or valgus were excluded. The participants underwent a gradual 2-week adjustment period with the FOD. During the first week they wore the device for progressive hour increments during daily activity and walking only. During the second week they were allowed to run with the orthoses in place for progressive increments of running mileage. Following the adjustment period, the participants returned to the lab for data collection of 3D kinematic, tibial acceleration, and GRF data on their right leg.

Retroreflective markers were placed over the greater trochanters bilaterally, on the medial and lateral femoral condyles, the medial and lateral malleoli, the forefoot, and on the medial and lateral borders of the first and fifth metatarsal head, respectively, in order to establish an anatomical coordinate system. Additionally, noncollinear tracking markers were placed on the pelvis, thigh, leg, and calcaneus (Figure 1). Following a standing calibration trial collected for each condition, the anatomical markers were removed.

Five trials of four conditions were collected: (a) RFS without the FOD; (b) RFS with the FOD; (c) FFS without the FOD; and (d) FFS with the FOD. All participants ran in Nike Air Pegasus shoes (Nike, Beaverton, OR). Holes were cut in the heel counter to allow the markers to be placed directly on the calcaneus (Figure 1). Previous testing was conducted to determine the optimal size of the holes such that the markers could project through the shoe without interference during running. Additional tests with an Instron device (Instron Corp., Canton,

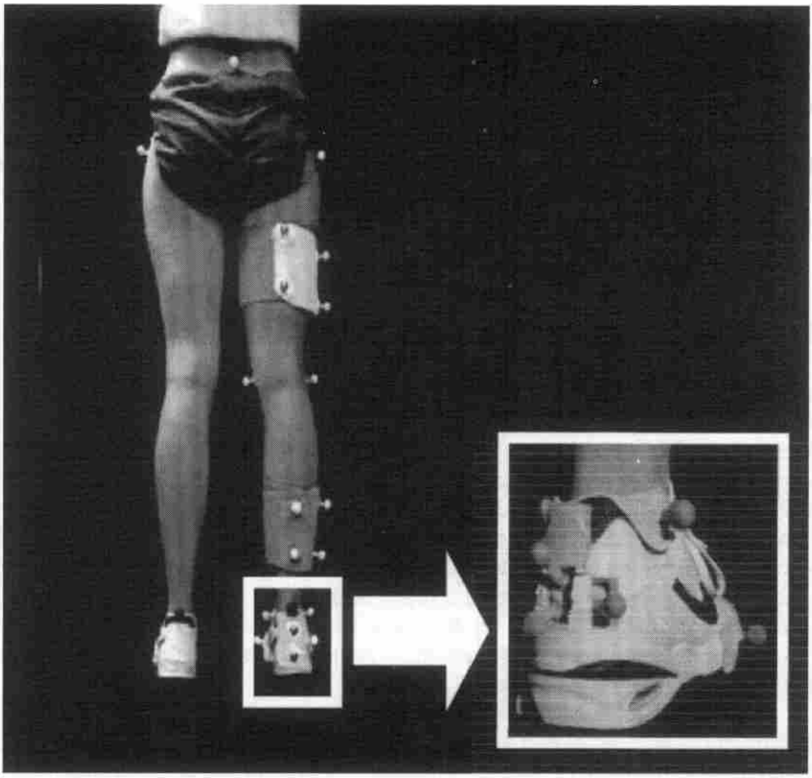

Figure 1 — Marker placement for kinematic and kinetic analysis. Inset: holes cut in the shoe to allow for placing markers directly on the calcaneus.

MA) revealed that these holes resulted in only a 10% decrement in heel counter stability. Both the FOD and strike pattern conditions were randomized for each runner to minimize order effects. For the FFS conditions, the runners were instructed to land on the ball of their foot and not let their heel contact the ground. Several practice trials were undertaken with the FFS pattern to allow the runners to adjust before any data were collected. Running speed averaged 3.7 $m \cdot s^{-1} \pm 5\%$ for both the RFS and FFS pattern, as determined by two photoelectric monotoring beams placed 285.5 cm apart.

A uniaxial accelerometer (model 353B17) and a sensor signal conditioner (model 480E09 ICD) set to a gain of 10 (PCB Pieziotronics, Depew, NY) were used to measure tibial accelerations. The device was protected in a custom machined aluminum casing and then mounted on a piece of lightweight thermoplastic material. The total weight of the mounted accelerometer was 3.28 g. It had a measurement range of $\pm$ 500 g (1 g = 9.8 $m \cdot s^{-2}$) and a mounted resonant frequency >70 kHz. With the gain set to 10, the voltage sensitivity of the acceleration signal was

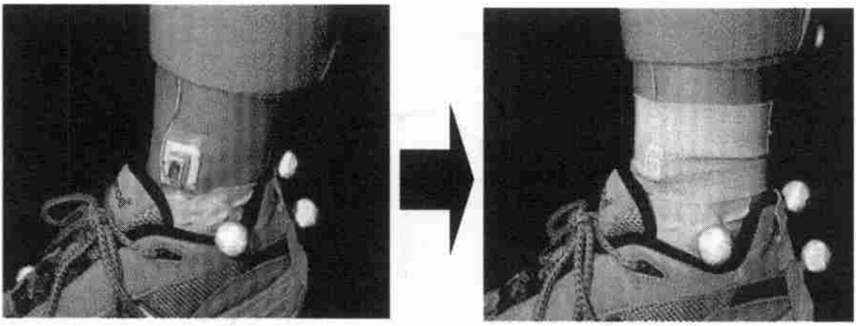

Figure 2 — Accelerometer placement upon distal anteriomedial aspect of the leg, fastened with double-sided tape and secured with Elastikon tape.

10mV/g, allowing for good sensitivity of the signal. The accelerometer was fastened to the distal anteriomedial aspect of the tibia leg in alignment with its longitudinal axis. Care was taken to place it away from the tongue of the shoe so as to prevent any interference during running (Figure 2). The distal location was chosen to reduce the effects of angular accelerations resulting from rotational movement about the ankle (Lafortune & Hennig, 1991). Elastikon (Johnson and Johnson, Arlington, TX) tape was tightly fastened over the accelerometer and around the leg to prevent the overestimation of peak positive acceleration due to skin movement (Hennig & Lafortune, 1988). The signal conditioner was fastened snugly around the runner's waist with a belt to minimize movement of the unit during each trial. The acceleration signal was sent to a 64-channel 12-bit A/D board (Vicon Motion Systems, Lake Forest, CA) via a coaxial cable 7.6 meters long.

Video, ground reaction force, and accelerometry data were collected simultaneously with a Vicon motion analysis system. Ground reaction force data were collected with a 60 × 90-cm Bertec force plate (Bertec Corp., Columbus, OH). Both accelerometery and GRF data were sampled at 960 Hz. Video data were collected at 120 Hz by six cameras placed at standardized locations and heights, with four cameras on the collection side and two cameras on the opposite side.

The 3D marker coordinates were low-pass filtered at 8 Hz using a second-order recursive Butterworth filter. A vertical GRF threshold of 10 N was used to determine heel strike and toe-off. All GRF data were low-pass filtered at 50 Hz with a second-order recursive Butterworth filter. This was done to decrease noise in the center-of-pressure data in early stance, improving the accuracy of the strike index (Cavanagh & Lafortune, 1980). The strike index was determined in order to validate the RFS and FFS trials. The strike index is a ratio of the distance of center of pressure from the heel, relative to foot length at the instant of initial foot contact (strike index for RFS trials is <0.33, for FFS trials it is >0.67).

Prior to data analysis, the acceleration signals were low-pass filtered at 100 Hz with a second-order recursive Butterworth filter. A power spectrum analysis revealed that 95% of the signal power of the acceleration signal was contained below 100 Hz. The average value and any linear trend in the acceleration signal

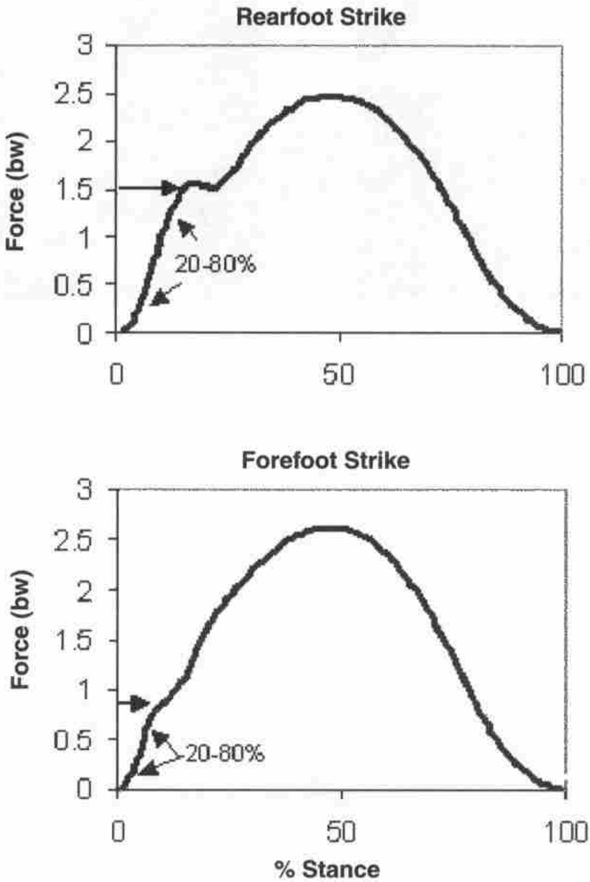

Figure 3 — Representative vertical ground-reaction-force load rates for a rearfoot strike and forefoot strike pattern. The rearfoot strike (RFS) average vertical load rate (VLR) was determined as 20–80% of slope to the first peak vertical GRF. The forefoot strike (FFS) average VLR was determined as 20–80% of slope to where the first change in slope occurred.

were removed, as described by Shorten and Winslow (1992). Peak positive acceleration was determined as the highest acceleration measurement of the time-series acceleration curve. PTP was determined as the total range of acceleration values during the stance phase.

The peak vertical ground reaction forces for the FFS and RFS patterns were determined. Average vertical GRF load rate in the RFS pattern was assessed from the slope (from 20 to 80% of the rise to the first impact peak) of the GRF (Figure 3). There was also present in the FFS pattern a small impact peak, or shift in the vertical GRF slope, which typically occurred earlier in the stance phase compared to the RFS pattern. Average vertical GRF load rate in the FFS pattern was also determined as 20 to 80% of the slope to this first peak or change in the GRF slope (Figure 3). Instantaneous vertical GRF load rate was determined as the peak change in force, as assessed by the central difference method over the same interval for which average GRF load rate was determined. The peak shear (anteroposterior) GRF and average anteroposterior GRF loading rates were also assessed for the FFS and the RFS patterns.

Rearfoot and knee joint kinematics and kinetics were calculated using Move3D (National Institutes of Health Biomechanics Lab, Bethesda, MD). All angular displacement data were resolved about a joint coordinate system (Grood & Suntay, 1983). Ankle dorsiflexion and knee flexion were positive while plantarflexion and knee extension were negative. Eversion excursion, dorsiflexion excursion, and knee flexion excursion were determined as the total angular range of motion in that plane during the first 60% of stance phase. Discrete variables were extracted from each of the runner's five trials and then averaged.

Ankle, knee, and leg stiffness as well as center-of-mass excursion were measured in order to provide additional information on the mechanisms of shock attenuation. Center-of-mass movement was determined from the second integral of the vertical GRF data (Cavagna, 1985). The total excursion of the center of mass was then assessed. Ankle and knee joint stiffness were calculated by modeling the ankle and knee as a rotational mass-spring (Hamill, Derrick, & McClay, 2000; Stefanyshyn & Nigg, 1998). This model derives stiffness from the slope of the torque-angle profile from the beginning of the stance phase to the peak joint moment during stance. Analog and video data were synchronized prior to calculation of joint kinetics by down-sampling the analog data to 120 Hz. Dempster's anthropometric data were used in the joint moment calculations (Dempster, 1990). Joint moments were standardized to participant height and body weight. Leg stiffness was estimated using a mathematical model presented by McMahon and Cheng (1990), which models the leg as a linear mass-spring.

Two-way repeated-measures analyses of variance (ANOVA) tests were conducted in order to test the independent variables of strike pattern and orthotic use on the tibial acceleration variables including peak positive tibial acceleration and peak positive to peak negative acceleration. Two-way ANOVA tests were also conducted to assess rearfoot eversion excursion, dorsiflexion excursion, knee flexion excursion, ankle stiffness, knee stiffness, leg stiffness, and center-of-mass excursion in order to further explain results. Significance was determined at $p < 0.05$.

Two bivariate regression analyses were conducted to assess the relationship between eversion excursion and peak positive tibial acceleration in an RFS and FFS pattern. Six additional bivariate regression analyses were conducted to assess the correlation between: (a) instantaneous vertical GRF load rate and peak positive tibial acceleration; (b) average vertical GRF load rate and peak positive tibial acceleration; and (c) average anteroposterior GRF load rate and peak positive tibial acceleration in an RFS and an FFS pattern. Student *t*-tests were used to assess differences in the ground reaction force variables of peak vertical GRF, average vertical GRF load rate, instantaneous vertical GRF load rate, peak anteroposterior GRF (peak braking force), and average anteroposterior GRF load rate between the FFS no-orthotic and the RFS no-orthotic conditions. Significance was determined at $p < 0.05$.

Table 1 Orthotic Condition and Strike Pattern Means (± *SD*) and Statistical Results

Variable	No FOD		FOD		FFS		RFS		*P*	*P*
	M	*SD*	*M*	*SD*	*M*	*SD*	*M*	*SD*	Ortho	Strike
Peak positive tibial acceleration (g)	7.18	(2.98)	6.78	(3.14)	7.82	(3.16)	6.15	(2.96)	0.231	*0.034*
Peak pos. to peak neg. acceler. (g)	12.76	(4.50)	13.04	(4.44)	14.90	(4.20)	10.88	(3.76)	0.660	*0.003*
Vertical GRF load rate (bw/sec)	55.77	(12.38)	51.81	(11.85)	52.94	(12.88)	54.64	(11.36)	*0.025*	0.99
Rearfoot eversion excursion (°)	15.59	(4.50)	14.45	(3.52)	16.38	(3.94)	13.66	(4.08)	0.105	*0.005*
Dorsiflexion excursion (°)	24.30	(4.87)	26.51	(4.21)	31.57	(4.31)	19.24	(4.77)	*0.017*	*<0.001*
Knee flexion excursion (°)	33.51	(3.29)	31.74	(3.54)	30.54	(3.84)	34.71	(2.99)	*0.037*	*<0.001*
Ankle stiffness(N·m/deg)	10.98	(3.02)	10.35	(2.54)	7.12	(2.03)	14.21	(3.53)	0.13	*<0.001*
Knee stiffness (N·m/deg)	5.00	(1.20)	5.24	(1.14)	5.47	(1.30)	4.77	(1.04)	*0.01*	*<0.001*
Leg stiffness (kN/m)	7.61	(1.26)	7.81	(1.11)	8.49	(1.41)	6.93	(0.96)	0.34	*<0.001*
Center-of-mass excursion (cm)	6.78	(0.53)	6.60	(0.64)	6.55	(0.50)	6.83	(0.67)	0.22	*0.02*

Note: Two-way repeated-measures ANOVA tests were conducted to test the independent variables of strike pattern and orthotic use on the tibial acceleration variables including peak positive tibial acceleration; and peak-positive to peak-negative acceleration. Two-way ANOVA tests were also conducted to assess average vertical ground reaction force load rate; rearfoot eversion excursion; dorsiflexion excursion; knee flexion excursion; ankle stiffness; knee stiffness; leg stiffness; and center-of-mass excursion to further explain results.
Italics indicate significance, $p < 0.05$

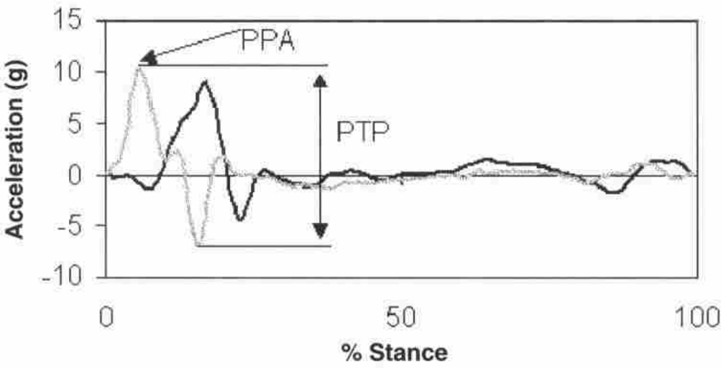

Figure 4 — Representative acceleration curves for a runner's RFS and FFS pattern. PPA = peak positive acceleration; PTP = peak-to-peak acceleration. Note the greater PPA and PTP in the FFS pattern. RFS acceleration data is represented by a solid black line, FFS acceleration data by a grey line.

Results

The FFS pattern exhibited a significant increase of 1.67 g in peak positive acceleration and a significantly greater peak-to-peak acceleration compared to the RFS pattern (Table 1; representative tibial acceleration signals are shown in Figure 4). The average peak vertical ground reaction force, the anteroposterior peak GRF, and average anteroposterior GRF load rates were significantly greater for the FFS pattern than for the RFS pattern (Table 2). Average and instantaneous vertical GRF load rate, however, did not differ significantly between the FFS and RFS patterns. The correlations between average vertical GRF load rate and peak positive acceleration for both FFS and RFS patterns were $r = 0.70$ and $r = 0.47$, respectively. Correlations between instantaneous load rate and peak positive acceleration in the FFS and RFS patterns were $r = 0.73$ and $r = 0.70$, respectively. Correlations of $r = 0.58$ and $r = 0.33$ were found between peak positive tibial acceleration and anteroposterior GRF load rate. All correlations were found to be significant ($P < 0.05$). On average, the values for the strike index were 0.80 ± 0.072 for the FFS conditions and 0.043 ± 0.048 for the RFS conditions, indicating a clear delineation between strike patterns.

Eversion excursion and dorsiflexion excursion were significantly greater for the FFS pattern compared to the RFS pattern, due to greater amounts of inversion and plantarflexion at foot strike with the FFS pattern. Knee flexion excursion was significantly lower in the FFS pattern, due to increased knee flexion at foot strike and decreased peak knee flexion compared to the RFS pattern. The FFS exhibited lower ankle stiffness than the RFS pattern, but greater knee and leg stiffness with associated decreases in center-of-mass excursion (representative ankle and knee stiffness graphs for RFS and FFS patterns are shown in Figure 5).

We did not find any significant differences between orthotic conditions for tibial accelerations or for rearfoot eversion excursion (Table 1). Thus there were no interaction effects between strike pattern and orthotic use for these parameters.

Table 2 GRF Variable Means (± *SD*) for Forefoot Strike and Rearfoot Strike Pattern With No Orthotics

Variable	FFS *M*	FFS *SD*	RFS *M*	RFS *SD*	*p*-value
Peak vertical ground reaction force	2.64	(0.18)	2.48	(0.13)	0.0002
Avg vertical GRF load rate	52.93	(15.00)	58.61	(9.77)	0.19
Instantaneous vertical GRF load rate	72.22	(16.60)	71.22	(12.19)	0.84
Peak anteroposterior GRF	0.47	(0.11)	0.367	(0.05)	0.002
Anteroposterior GRF load rate	26.17	(8.72)	9.46	(3.39)	<0.0001

Note: Units for ground reaction forces are in body weights; Units for loading rates are in body weights per second.

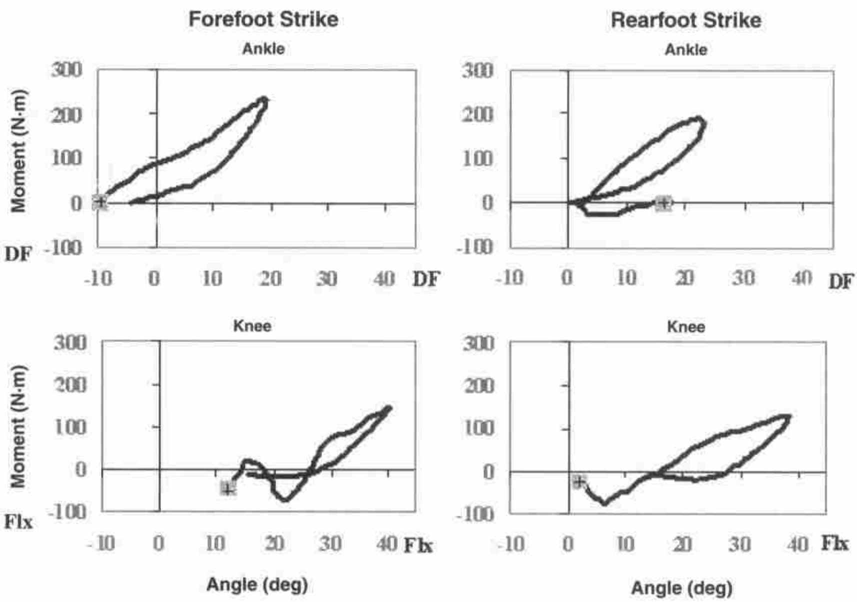

Figure 5 — Representative ankle and knee stiffness diagrams for FFS and RFS from a single runner. Stiffness was calculated as the average slope of the moment angle profile during the loading phase of stance (heel strike to peak angle and moment). A steeper positive slope indicates greater stiffness. Note the greater ankle stiffness in RFS and the greater knee stiffness in FFS. Foot strike is indicated by ⊞. Ankle plantarflexion moment is positive, ankle dorsiflexor (DF) moment is negative, plantarflexion angle is negative, dorsiflexion angle (DF) is positive. Knee extensor moment is positive, knee flexor moment (Flx) is negative, knee extension angle is negative, knee flexion angle (Flx) is positive.

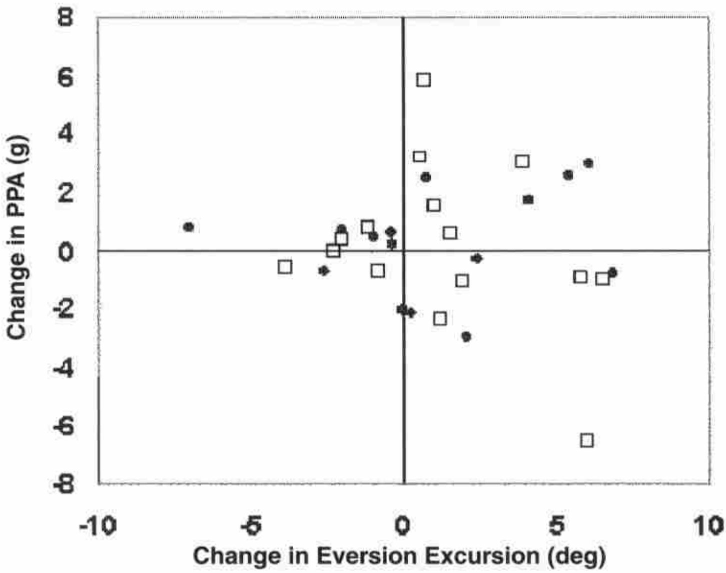

Figure 6 — Change in rearfoot eversion excursion vs. change in peak positive acceleration (PPA) from barefoot to FOD condition in both the RFS and FFS patterns. A negative value indicates a reduction in the variable as a result of the orthoses. The correlation between the change in eversion excursion and change in PPA for the RFS was $r = 0.21$; for FFS it was $r = -0.30$. RFS data represented by filled circles, FFS data by open squares.

Dorsiflexion excursion was significantly greater in the orthotic condition as a result of decreased dorsiflexion at foot strike with similar peak dorsiflexion angles. Average vertical GRF load rate was significantly reduced in the orthotic condition. The orthotic intervention resulted in similar values for leg stiffness, ankle stiffness, and center-of-mass excursion compared to the no-orthotic condition. Significantly greater knee stiffness values were observed in the orthotic condition compared to the no-orthotic condition. Knee flexion excursion was significantly decreased with orthotic intervention as a result of increased knee flexion at foot strike with similar peak knee flexion. There was no relationship between the change in eversion excursion and the change in peak positive acceleration in either the RFS or FFS pattern between the orthotic and the no-orthotic condition (Figure 6).

Discussion

The purpose of this study was to investigate the effects of an RFS vs. FFS pattern and orthotic intervention on shock to the lower extremity. Kinematic values associated with each strike pattern fell within the ranges reported in the literature (DeWit, DeClercq, & Aerts, 2000; McClay & Manal, 1995a; Smith, Clarke, Hamill, & Santopietro, 1986; Valiant, McMahon, & Frederick, 1987; Viale, Belli, Lacour, & Freychat, 1997; Williams et al., 2000), as did kinetic (McClay & Manal, 1995b) and GRF values (Hennig et al., 1993; McClay & Manal, 1995b). In addition, the mean peak positive acceleration for the RFS data were comparable to those reported in other studies (Clarke, Cooper, Hamill, & Clarke, 1985; Hennig & Lafortune, 1993; Lafortune & Hennig, 1991; Lafortune, Hennig, & Lake, 1996; Lafortune, Hennig, & Valiant, 1995; Mahar, Derrick, Hamill, & Caldwell, 1997; McMahon, Valiant, & Frederick, 1987; Oakley & Pratt, 1988).

Increased tibial accelerations were found in the FFS pattern along with increased peak vertical and anteroposterior GRF, greater anteroposterior GRF load rates, greater leg stiffness, and greater knee stiffness. Orthotic intervention did not significantly change tibial shock or rearfoot eversion excursion, but it did have an effect on dorsiflexion excursion, knee flexion excursion, and knee stiffness.

All of the acceleration patterns consisted of an initial positive acceleration, followed by a negative acceleration within the first third of stance phase. This pattern is characteristic of several other studies (Hennig & Lafortune, 1991; Hennig et al., 1993; Lafortune & Hennig, 1991; Lafortune et al., 1995; Mahar et al., 1997). The negative value may be due in part to the accelerometer wobbling relative to the bone. However, this type of artifact was minimized due to the lightweight design of the accelerometer and its tight fixation to the skin above the tibia. This negative value may also result from the angular motion of the tibia. Lafortune and Hennig (1991) found that angular motion contributes approximately –5.06 g to the acceleration signal during running. We placed the accelerometer on the distal tibia in order to minimize the effects of angular motion on the acceleration signal; however, angular motion of the ankle did occur, as indicated by the ankle dorsiflexion excursion values.

Since it was predicted that the vertical GRF load rate would be significantly greater in the RFS pattern than in the FFS pattern, we expected tibial shock to be greater in an RFS compared to an FFS pattern. This hypothesis was based on a previous study by Hennig and co-workers (Hennig & Lafortune, 1991; Hennig et al., 1993) in which a strong positive correlation between vertical GRF load rate and peak positive acceleration in RFS was reported. Significant positive correlations were found in the present study between vertical GRF load rates and PPA in both the RFS and FFS patterns. However, no significant differences between vertical GRF load rates were found between strike patterns. Hennig and Lafortune (1991) also found peak anteroposterior GRF to be a significant predictor of PPA. In this study, both peak and average anteroposterior GRF load rates were significantly greater in the FFS running pattern. In addition, the correlation between these shear load rates and peak positive acceleration was significant. Shear forces imparted along the longitudinal axis of the foot may be transmitted to the longitudinal axis of the tibia through the subtalar joint, thus applying a load to the tibia.

The greater peak acceleration in the FFS may also be related to peak forces. On average, the FFS pattern resulted in greater vertical GRF peaks by 16% of body weight, or an average force difference of 104 Newtons. Previous studies have also reported greater peak vertical GRFs with an FFS pattern (McClay & Manal, 1995b; Oakley & Pratt, 1988). This difference was significant and most likely contributed to the greater amount of tibial acceleration in the FFS pattern. Finally, it is important to note that the heel pad of the foot provides an additional mechanism of shock attenuation in the RFS vs. the FFS. The smaller peak acceleration values of the RFS and greater vertical GRF force peaks in the FFS pattern may be due, in part, to the ability of the heel pad to absorb a substantial amount of energy (Valiant & Cavanagh, 1985).

To further explain the greater tibial shock in the FFS pattern, we also assessed leg, knee, and ankle stiffness. Joint stiffness is a function of angular excursion and joint torque. Compared to the RFS, the FFS pattern demonstrated greater dorsiflexion excursion and therefore lower ankle stiffness. Knee flexion excursion was lower in the FFS pattern, however, resulting in greater knee stiffness compared to the RFS pattern. Leg stiffness was also greater in the FFS pattern, suggesting that the knee may be a stronger modulator of leg stiffness than the ankle. In a similar study of RFS and FFS patterns, Hamill et al. (2000) also noted the greater leg stiffness in an FFS pattern and the knee's greater contribution to leg stiffness. As further support of the knee's role in leg stiffness, Arampatzis, Bruggemann, and Metzler (1999) reported greater increases in knee and leg stiffness, and less substantial increases in ankle stiffness with increased running speeds.

The greater peak-to-peak acceleration seen in the FFS pattern compared to the RFS pattern is in contrast to a study by Oakley and Pratt (1988). These researchers, however, mounted their accelerometer to a large splint which was molded around the ankle and rested upon the medial and lateral malleoli, as opposed to our methodology in the present study of attaching the accelerometer on the skin superficial to the tibia. Additionally, while the weight of the combined accelerometer and splint was not reported, the accelerometer model used by Oakley and Pratt (1998) was 6.3 grams heavier than the one we used in the present study. Hennig and Lafortune (1988) report a 50% error when using a skin-mounted accelerometer package with a mass that exceeds 6 g. These factors may have affected the shock measurements between strike patterns.

A possible limitation of the present study was that natural FFS runners were not used. While Williams et al. (2000) had perviously established that converted FFS runners were similar in their mechanics to natural FFS runners, they did not address tibial acceleration measures. Therefore it is possible that the increase in tibial shock seen in the FFS pattern of the present study was an artifact of this novel skill. In addition, our runners were instructed to run with a toe-strike pattern and not allow their heel to contact the ground. This may have resulted in an artificial stiffening of the leg. While some natural forefoot strikers do remain on the ball of their foot during contact, others dorsiflex the ankle and make heel contact, resulting in increased dorsiflexion excursion. It is possible that, due to increased dorsiflexion excursions, this latter type of FFS runner would exhibit lower tibial shock compared to an FFS pattern with forefoot contact only.

Orthotic intervention did not alter tibial shock in either foot-strike pattern. The hypothesis was based on the idea that restriction of rearfoot motion would increase tibial shock. However, eversion excursion was not significantly reduced with the use of the FOD. In addition, as shown in Figure 6, there was no correlation ($R = 0.21$ and -0.30 for RFS and FFS patterns, respectively) between changes in eversion excursion and changes in peak positive acceleration with orthotic intervention in either the RFS or the FFS pattern. Although one runner did exhibit a 7-g increase in peak positive acceleration with a 5° decrease in eversion excursion with the orthosis, others with similar reductions in eversion excursion showed no change or decrease in peak positive acceleration. While rearfoot movement was not restricted by the orthosis, a significant increase in ankle dorsiflexion occurred in the orthotic conditions. This may be a compensation for restriction of joint motion of the midfoot with orthotic use. However, it is difficult to detect this restriction in midfoot motion with noninvasive 3D motion analysis. Previous studies have reported restriction of rearfoot motion with orthotic use in a patient population. The runners in the present study had normal lower extremity alignment and did not have running-related overuse injuries. It is possible that greater restriction of motion would have been seen in runners who needed more rearfoot motion control.

It was interesting to note that vertical load rate was significantly decreased with orthotic intervention. This may account for the reported reduction in both metatarsal and tibial stress fractures in military recruits with the use of FOD (Simkin, Leichter, Giladi, Stein, & Milgrom, 1989), as impulsive loading has been associated with stress fractures in bone studies (Boyce, Fyhrie, Glotkowski, Radin, & Schaffler, 1998; Burr et al., 1990). Given the strong relationship between vertical GRF load rate and peak positive acceleration reported by Hennig and colleagues (Hennig & Lafortune, 1991; Hennig et al., 1993), it was surprising that there was not an associated significant reduction of peak positive acceleration in the FOD condition. Hennig and Lafortune (1991) studied this relationship under an array of footwear conditions, but they did not examine it under different orthotic conditions or strike patterns. The average vertical and peak vertical GRF load rate values in the RFS data of the present study were similar to those reported by Hennig and colleagues. However, the correlations to peak positive acceleration were slightly lower.

In conclusion, FOD did not significantly change tibial acceleration for either an RFS or FFS pattern in a group of uninjured recreational runners. The FFS pattern, however, did result in greater tibial acceleration compared to the RFS pattern. In the FFS pattern, although ankle stiffness was lower, knee and leg stiffness was greater than in the RFS pattern. Peak vertical GRF, shear GRF load rates, and shear GRF peaks were greater for the FFS pattern, which may account for the increase in tibial acceleration. Future studies should examine the effects of orthotic intervention on natural FFS runners and compare tibial acceleration of natural FFS runners to that of natural RFS runners. Research on the effects of orthotic use on tibial acceleration in a patient population may shed light on the relationship between eversion excursion and tibial acceleration. In addition, tibial acceleration in FFS patterns that allow heel contact needs to be examined. Also, the effect of running speed on both tibial acceleration and rearfoot eversion warrants further investigation. A better understanding of the differences between FFS and RFS patterns may help in the identification of injury risks and development of preventative strategies for runners with varying strike patterns.

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