Effect of Inverted Orthoses on Lower-Extremity Mechanics in Runners

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ABSTRACT

WILLIAMS III, D. S., I. MCCLAY DAVIS, and S. P. BAITCH. Effect of Inverted Orthoses on Lower-Extremity Mechanics in Runners. *Med. Sci. Sports Exerc.*, Vol. 35, No. 12, pp. 2060–2068, 2003. **Introduction:** Foot orthoses are recommended for individuals with injuries associated with abnormal lower-extremity mechanics. However, the biomechanical effect of these devices is not completely understood. Most clinicians and researchers believe that foot orthoses are effective in reducing some aspect of rearfoot motion. This is important as many injuries are suggested to be the result of increased pronation. Inverted orthoses are a more aggressive treatment in those whose symptoms do not respond to standard orthotics. They are likely to alter motion in all planes. However, no three-dimensional studies have assessed lower-extremity mechanics in individuals wearing inverted orthotics. **Purpose:** The purpose of this study was to compare the three-dimensional kinematics and kinetics of the rearfoot and knee during running while varying orthotic intervention. **Methods:** Eleven subjects were initially fitted with standard foot orthoses and then with inverted orthoses. Three-dimensional kinematic and kinetic data were collected for conditions of no orthoses, standard orthoses, and inverted orthoses. **Results:** There were no differences between conditions in peak rearfoot eversion or rearfoot eversion excursion. Peak rearfoot inversion moment and work were significantly reduced ($P = 0.045$ and $P < 0.001$, respectively) in the inverted orthotic condition suggesting a decreased demand on the soft tissue structures that control eversion. Significant differences were seen in tibial rotation ($P = 0.043$), knee adduction ($P = 0.035$), and knee abduction moment ($P < 0.001$) in the inverted orthotic condition, suggesting alterations were made further up the kinetic chain. **Conclusions:** The differences in kinetic parameters at the rearfoot may result in fewer injuries of the rearfoot soft tissue structures when using inverted orthotics. These alterations in lower-extremity mechanics associated with inverted orthoses provide clinicians some evidence for prescribing this device. **Key Words:** KINETICS, KINEMATICS, INJURY

Foot orthoses are commonly recommended for individuals with running injuries associated with abnormal lower-extremity mechanics. Numerous studies have documented the clinical efficacy of these devices (7,10,14,17,27). A number of studies have investigated the effects of foot orthoses on the lower extremity. However, most studies have focused on the two-dimensional motion of the rearfoot and have reported mixed results (1,4,15,30,31). Three-dimensional studies of the rearfoot, tibia, and knee have also been reported (20–22,24,30,32). Although there are discrepancies in the literature, the majority of studies report that foot orthoses are effective in reducing some aspect of rearfoot motion including peak rearfoot eversion (1,2,20,22,26,30), rearfoot eversion excursion (23,26), and rearfoot eversion velocity (26,31). This is important as many lower-extremity injuries are suggested to be the result of excessive pronation (12), although the criteria for "excessive" are often not well defined.

Foot orthoses are also often prescribed for individuals with knee-related pathologies such as patellofemoral pain syndrome, iliotibial band syndrome, and meniscal pain. This is based on the idea that rearfoot and knee motions are coupled. This coupling has been shown to occur through the subtalar and navicular joints (16). As the foot pronates at both joints, the talus rotates internally. Due to the tight articulation in the ankle mortise, the tibia also rotates internally. McClay and Manal (19) studied the lower-extremity mechanics of runners with excessive rearfoot eversion. They defined excessive as having a peak two-dimensional eversion greater than 18°. This was based upon two-dimensional rearfoot mechanics previously collected on 100 normal runners whose mean rearfoot angle was 12° (SD = 3°). Eighteen degrees, therefore, places these runners outside 2 standard deviations of this sample population. These runners exhibited significantly greater knee flexion, abduction (valgus), and internal rotation than runners with normal rearfoot motion (8–12° eversion). These increased motions are thought to lead to abnormal patellofemoral joint alignment and result in patellofemoral joint pain, one of the most common injuries that runners sustain (6,12). The goal of the orthotic device in these cases is to indirectly

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TABLE 1. Subject characteristics.

Subject	Sex	Age	Mass (kg)	Involved Side	Diagnosis	RCSP (°)	Inversion angle of orthoses (°)
1	F	41	45.2	Left	PTT	12	25
2	M	49	100.0	Left	PF	14	25
3	M	23	64.1	Left	ACS	8	15
4	M	23	65.5	Left	ACS	7	15
5	M	30	94.1	Right	ACS	12	25
6	F	18	49.2	Right	ACS	11	25
7	F	20	68.7	Right	PFP	11	25
8	F	18	66.2	Right	PF	12	25
9	F	18	62.1	Right	PF	10	15
10	M	52	78.2	Right	PF	6	15
11	F	37	68.2	Left	PFP	8	15

RCSP, resting calcaneal stance position; F, female; M, male; PTT, posterior tibialis tendonitis; PF, plantar fasciitis; ACS, anterior compartment syndrome; PFP, patellofemoral pain syndrome.

realign the knee by altering the foot alignment. However, two additional studies (8,34) using a patient population found no changes in the knee mechanics with foot orthotic devices. It should be noted that, the study by Eng and Pierrynowski (8) used a flat Spenco® insole with forefoot and rearfoot wedges, which may not have controlled the foot as well as a custom device.

Inverted orthoses were developed as a more aggressive approach to treatment of runners whose symptoms do not respond to standard orthotic treatment (3). The positive mold of the inverted orthoses is poured in varying degrees of inversion as opposed to the vertical orientation of a standard orthoses. According to Blake (3), due to the soft tissue compression of the heel, in order to achieve a 1° change in the calcaneus, a 5° rearfoot varus post is needed. Thus, a 5° correction of the rearfoot would require a 25° inverted device. Blake et al. (3) compared the rearfoot mechanics of patients treated with both a standard and inverted orthoses. They found that the 25° inverted orthoses was more effective in controlling rearfoot pronation than a standard orthotic device (1). However, these authors used a two-dimensional approach, which becomes inaccurate as the lower extremity rotates in the transverse plane (18). Three-dimensional analysis of the effect of inverted orthoses on lower-extremity kinetics may provide a more comprehensive justification for the use of these devices in the successful treatment of lower-extremity pathology.

Therefore, the purpose of this study was to compare the effect of a standard and inverted orthotic device on the three-dimensional kinematics and kinetics of the rearfoot and knee in a group of runners successfully treated with the inverted orthotic device. It was hypothesized that the inverted orthoses would result in greater reduction of peak rearfoot eversion, eversion excursion, and associated inversion moment than either standard orthotic devices or no orthoses condition. Additionally, it was hypothesized that peak knee adduction angle and associated knee abduction moment would be increased and knee internal rotation would be decreased to a greater degree in the inverted orthoses condition than the standard orthoses condition when compared with the no orthoses condition.

METHODS

Eleven runners (5 males, 6 females) between the ages of 20 and 52 (mean = 30.6 ± 11.4 yr) volunteered for this study. All methods were approved by the Institutional Review Board at the University of Delaware. Written informed consent was obtained from all subjects before

FIGURE 1—Posterior view of orthotic devices. A left inverted orthosis is on the left and is characterized by a more aggressive lateral tilt of the entire device. The external posting is the white material on the heel of the right standard orthosis on the right.

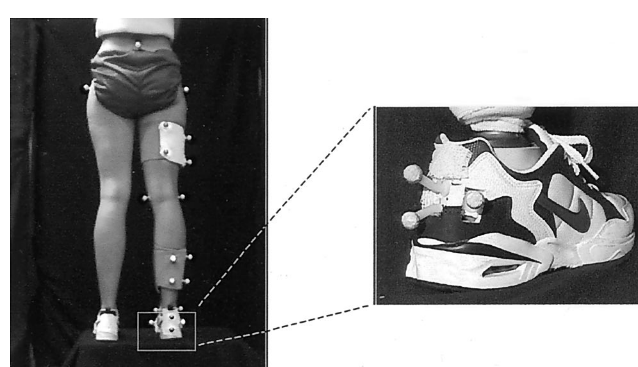

FIGURE 2—Retroreflective marker placement on the tested lower extremity. The enlargement shows placement of the rearfoot markers on the calcaneus projecting through holes in the shoe.

TABLE 2. Rearfoot kinematic and kinetic variables.

	IO	SO	NO	*P*	Effect Size
Eversion excursion (°)	15.76 (2.84)	14.99 (3.37)	15.81 (2.84)	0.476	0.01
Pk. eversion (°)	8.69 (3.38)	9.08 (3.81)	7.49 (3.95)	0.518	0.16
Pk. inv. moment ($N \cdot m \cdot kg^{-1} \cdot m^{-1}$)	−0.12 (0.10)	−0.19 (0.15)	−0.26 (0.16)	0.045*	0.46
Pk. power abs. ($W \cdot kg^{-1} \cdot m^{-1}$)	−0.34 (0.10)	−0.40 (0.33)	−0.52 (0.45)	0.303	0.27
Neg. work ($J \cdot kg^{-1} \cdot m^{-1}$)	−0.44 (0.37)	−1.07 (0.65)	−1.80 (1.04)	<0.001#	0.66

IO, inverted orthoses; SO, standard orthoses; NO, no orthoses; Pk., peak; inv., inversion; abs., absorption; Neg., negative.
* Tukey comparisons reveal a significant difference only between the IO and NO conditions.
\# Tukey comparisons reveal significant differences between IO and SO, SO and NO, and IO and NO conditions.

participation in the study. These subjects had initially been fitted with standard orthoses to relieve a variety of running-related injuries (Table 1). Each standard orthotic device was custom molded and internally posted to the forefoot deformity. Further, each pair of orthoses was made with a graphite shell and externally posted in 4° of rearfoot varus. The intrinsic post allows the forefoot to be supported in its natural position when the rearfoot is in a neutral position. The external post supports the rearfoot at the subtalar neutral position and further adds a standard 4° varus correction. In response to only partial or limited relief of symptoms, these patients were then fitted with graphite inverted orthoses posted between 15° and 25°, depending upon the severity of their structure and symptoms (Fig. 1). The degree of posting is based on typical clinical prescription of the inverted orthotic devices. Subjects with a resting calcaneal stance position of 5–10° were posted at 15° while those with a stance position of greater than 10° were posted at 25° (Table 1). All subjects were able to return to pain-free running with the use of the inverted orthoses.

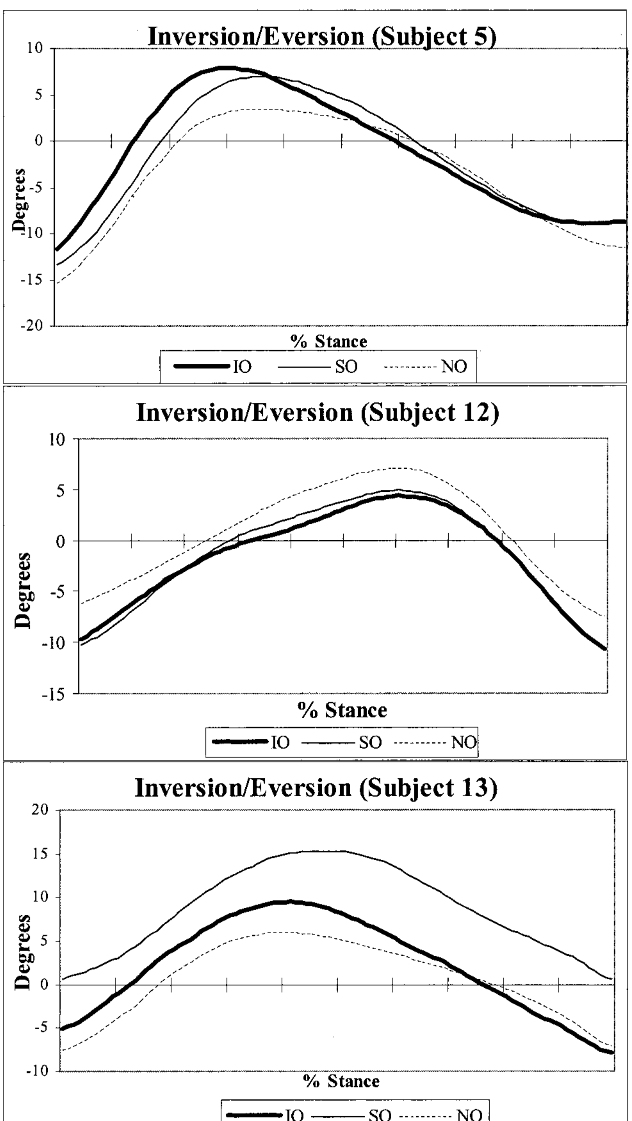

FIGURE 3—Frontal plane rearfoot motion in three selected subjects. Eversion is positive and inversion is negative. Subject 5 demonstrates the opposite of the expected effect with eversion increasing as orthotic correction increases. Subject 12 demonstrates the expected results whereas subject 13 presents with a random effect.

A three-dimensional lower-extremity gait analysis focusing on joint motion and joint kinetics was performed on these subjects at the Motion Analysis Laboratory. Retroreflective markers were placed unilaterally on the segments of the rearfoot, shank, thigh, and pelvis of the affected or most symptomatic side (Fig. 2). Anatomical markers were placed over bilateral greater trochanters, medial and lateral femoral condyles, medial and lateral malleoli, medial and lateral forefoot, and at the most anterior point on the end of the shoe. Four tracking markers were placed on the thigh and shank, and three were placed on the pelvis and rearfoot. The rearfoot markers were placed directly on the heel and extended through windows cut in the shoes. Windows allowed for unabated motion of the markers on the heel. Using an Instron materials testing device (Canton, MA), a standard force of 20 lb was placed on the lateral border of the heel counter of each shoe, and linear displacement was measured and compared with and without the heel cutout. The cutouts only result in approximately 10% decrement in heel counter stability.

An anatomical coordinate system was established for each of the foot, shank, thigh, and pelvis in order to determine joint kinematics and kinetics (35). Kinematic data were collected at 120 Hz using a six-camera VICON motion analysis system (Oxford Metrics Limited, UK). A force plate (BERTEC, Worthington, OH) mounted in the center of a 25-m runway recorded ground reaction forces at 480 Hz. Anatomical markers were then removed after a standing calibration. All subjects wore the same brand and model of shoes in order to reduce variability related to footwear. The

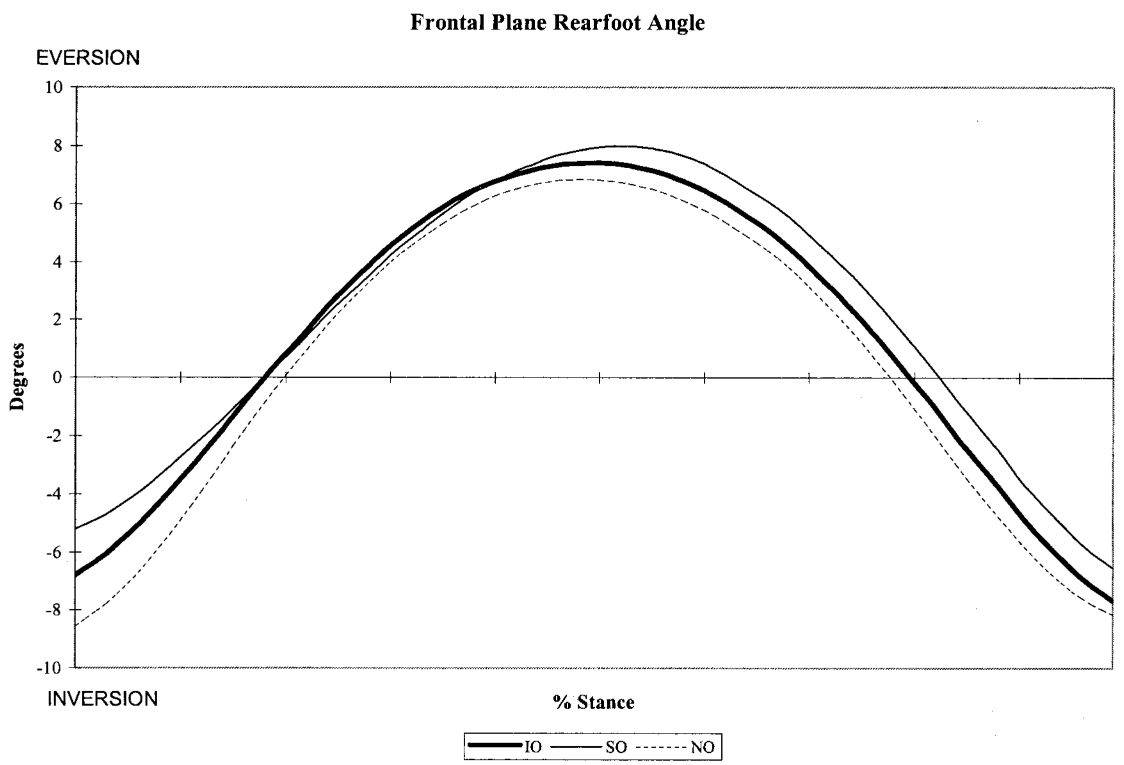

FIGURE 4—Group means for frontal plane rearfoot motion. All three groups demonstrate approximately the same eversion excursion. Note that the SO group reaches a larger degree of peak eversion than both the IO and NO groups. Standard deviations are not presented for clarity of means.

subjects ran along the runway at a speed of 3.35 $m \cdot s^{-1}$ (8 $min \cdot mile^{-1}$ pace). This was a comfortable pace for all subjects, as it was close to their training pace. All subjects were given adequate opportunity to accommodate to running on the runway with the markers in place. Speed was monitored with photocells, and only trials within ±10% of the target speed were accepted. Five trials were collected for each of the randomized conditions as subjects ran without an orthoses, with their standard orthoses and with their inverted orthotic devices.

All data were analyzed between heel strike and toe off and normalized to 100 data points. The three-dimensional coordinates of each marker were reconstructed using the VICON motion analysis software (Oxford Metrics Limited, UK). The three-dimensional coordinates were filtered using a second order recursive Butterworth filter with an 8-Hz low-pass cutoff frequency. Force data were low pass filtered at 50 Hz. MOVE3D software (National Institutes of Health Biomechanics Laboratory, Bethesda, MD) was used to determine joint kinematic and kinetic data (11). Joint angles were resolved about a joint coordinate system (9). Joint moments and powers were normalized to body mass and height. Means of discrete variables (i.e., peak eversion) were calculated using each individual's peak. Mean graphs were calculated from individual graphs and may not reflect similar values, as reported peak values across individuals likely do not occur at the same time.

TABLE 3. Individual subjects data on selected variables.

	RF Peak Eversion			RF Eversion Excursion		
	NO	SO	IO	NO	SO	IO
1	5.7	−1.4	*7.4*	13.7	12.1	12.1
2	5.1	*11.5*	*8.5*	16.4	12.9	14.4
3	3.5	*9.7*	*7.2*	19.0	*22.2*	*21.9*
4	16.8	10.7	11.8	20.4	18.0	19.7
5	4.2	*7.5*	3.4	11.7	*16.4*	*13.6*
6	9.2	*14.45*	7.0	17.2	14.0	14.8
7	13.4	10.6	11.4	19.7	16.6	17.0
8	4.5	*11.6*	*15.0*	16.1	13.9	*17.7*
9	6.7	*7.6*	*12.5*	12.9	8.4	*13.6*
10	7.2	*9.16*	7.0	13.3	*15.1*	*14.0*
11	6.2	*8.5*	4.6	13.8	*15.4*	*14.7*
Mean	7.49	9.08	8.69	15.81	14.99	15.76
SD	3.95	3.82	3.38	2.84	3.37	2.84

IO, inverted orthoses; SO, standard orthoses; NO, no orthoses.
Italics represent an increase compared with the NO condition.

Both rearfoot eversion excursion and peak eversion were evaluated in the current study. Eversion excursion is the total range of motion that the rearfoot traverses from heel strike to peak eversion. Peak eversion occurs at or near midstance. Rearfoot inversion peak power and negative work were also evaluated during the first half of stance in order to estimate muscular demands at the rearfoot during the eccentric phase of loading. Joint power was estimated using an inverse dynamic approach and work was calculated as the integral of the power curve during the first 50% of stance. Peak tibial and knee internal rotation, knee adduction, and peak abduction moment were also compared in order to demonstrate changes occurring further up the lower kinetic chain. Comparisons between orthotic conditions were made using a repeated measures anal-

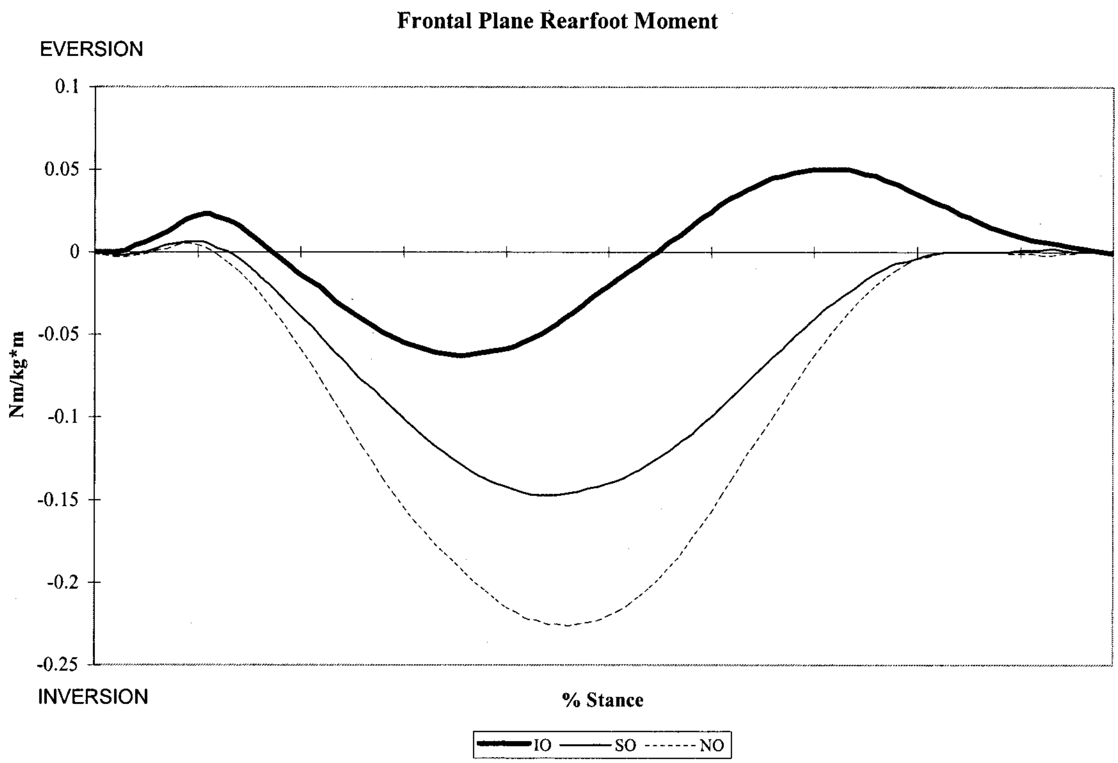

FIGURE 5—Group means for rearfoot inversion/eversion moment curves demonstrate a greatly reduced inversion moment during the first 50% of stance. Note the eversion moment late in stance in the IO condition. Standard deviations are not presented for clarity of means.

ysis of variance ($P \leq 0.05$) to determine whether differences exist in rearfoot eversion excursion, peak eversion, and peak inversion moment. Once significant differences were determined in the ANOVA, Tukey *post hoc* tests were employed to determine specific differences between pairs of the three groups.

RESULTS

No significant difference was noted between conditions in rearfoot eversion peak or excursion (Table 2). The frontal plane kinematic response to the inverted orthoses at the rearfoot was quite variable (Fig. 3), which may explain in part why there was no appreciable difference in the rearfoot motion curves (Fig. 4). Surprisingly, 5 of 11 and 6 of 11 individuals demonstrated an increase in peak eversion and eversion excursion, respectively, between no orthoses and inverted orthoses conditions (Table 3).

There was, however, a systematic decrease in inversion moment between the conditions of no orthoses, standard orthoses, and inverted orthoses. During the inverted orthoses condition, there was an eversion moment late in stance that was absent in the no orthoses and standard orthoses conditions (Fig. 5). Peak rearfoot inversion moment was significantly decreased in the inverted orthoses condition when compared with no orthoses condition at the rearfoot (Table 2). Peak inversion moment was significantly decreased by 54% between the no orthoses and inverted orthoses conditions, with 10 of 11 subjects demonstrating this pattern. There was a 27% decrease in the inversion moment between the no orthoses and standard orthoses conditions, with 8 of 11 subjects showing this trend; however, this decrease was not significant (Table 4). Although the absolute differences in peak inversion moment were small between groups, effect sizes show a moderate effect of 0.39 between the inverted and standard orthoses groups and 0.46 between the inverted and no orthoses groups.

There was no significant difference between groups in peak power absorption, whereas there was a systematic decrease the negative work in the frontal plane of the rearfoot across the three conditions. There was a 40.6% decrease between the no orthoses and standard orthoses conditions, and a 75.6% difference between the no orthoses

TABLE 4. Individual subjects data on rearfoot peak inversion moment.

	NO	SO	IO
1	−0.18	*−0.12*	*−0.13*
2	−0.13	−0.20	*−0.00*
3	−0.12	−0.16	*−0.02*
4	−0.20	*−0.05*	*−0.10*
5	−0.09	*−0.00*	*0.00*
6	−0.66	*−0.60*	*−0.00*
7	−0.37	*−0.27*	*−0.22*
8	−0.31	−0.32	*−0.08*
9	−0.21	*−0.08*	*−0.30*
10	−0.21	*−0.15*	*−0.18*
11	−0.34	*−0.18*	*−0.27*
Mean	−0.257	−0.193	−0.117
SD	0.155	0.154	0.104

IO, inverted orthoses; SO, standard orthoses; NO, no orthoses.
Italics represent a decrease compared with the NO condition.

TABLE 5. Knee kinematic and kinetic variables.

	IO	SO	NO	P	Effect Size
Pk. ADD (°)	8.61 (3.69)	7.05 (4.26)	5.10 (4.65)	0.035*	0.39
Pk. ABD moment ($N \cdot m \cdot kg^{-1} \cdot m^{-1}$)	−0.62 (0.15)	−0.58 (0.16)	−0.49 (0.16)	<0.001*	0.39
Pk. tibial int rot (°)	10.79 (3.73)	9.62 (6.41)	6.29 (6.19)	0.043*	0.40
Pk. knee int rot (°)	−0.98 (3.2)	0.33 (4.17)	−0.54 (3.86)	0.392	0.06

IO, inverted orthoses; SO, standard orthoses; NO, no orthoses; Pk., peak; ADD, adduction; ABD, abduction; int rot, internal rotation.
* Tukey comparisons reveal a significant difference only between the IO and NO conditions.

and inverted orthoses conditions when comparing negative work with these findings being statistically significant.

Knee kinetic and kinematic data in the frontal plane were also significantly different between conditions (Table 5). Peak knee adduction increased as orthotic correction increased from no orthoses to inverted orthoses (Fig. 6). There was an associated progressive increase in peak knee abduction moment across all three conditions (Fig. 7). There were no differences across conditions in knee internal rotation (tibial motion relative to femoral motion). However, internal rotation in the tibia (relative to fixed foot segment) was significantly increased in the inverted orthoses condition when compared with the no orthoses condition (Table 5).

DISCUSSION

The purpose of this study was to compare the effect of a standard and inverted orthotic device on the lower-extremity mechanics of a group of successfully treated injured runners. Contrary to the hypotheses, there were no significant differences in rearfoot eversion among all orthotic conditions. Other researchers have reported similar findings (4,13,27,30). Recent findings also suggest a varied response in rearfoot kinematics to orthotic devices (24,25,32). This lack of finding may be related, in part, to the large variability in response to the orthoses. Some subjects showed as much as a 10° increase and others a 5° decrease in peak rearfoot eversion when comparing the inverted orthoses condition with the no orthoses condition. This variability in response to orthoses has been reported elsewhere (28).

It was interesting to note that approximately half of the subjects actually increased their peak eversion and eversion excursion when using the inverted devices. Nawoczenski et al. (22) found that individuals with flat feet had increased eversion excursion and peak when wearing orthoses. Although foot structure was not assessed in the subjects of the present study, all of the runners had been prescribed orthoses for pain related to overpronation, which is often associated with pes planus (33). This may have accounted for some of the subjects demonstrating an increase in eversion in the orthotic conditions. All subjects were regular wearers of inverted orthoses. One limitation of this study was that it

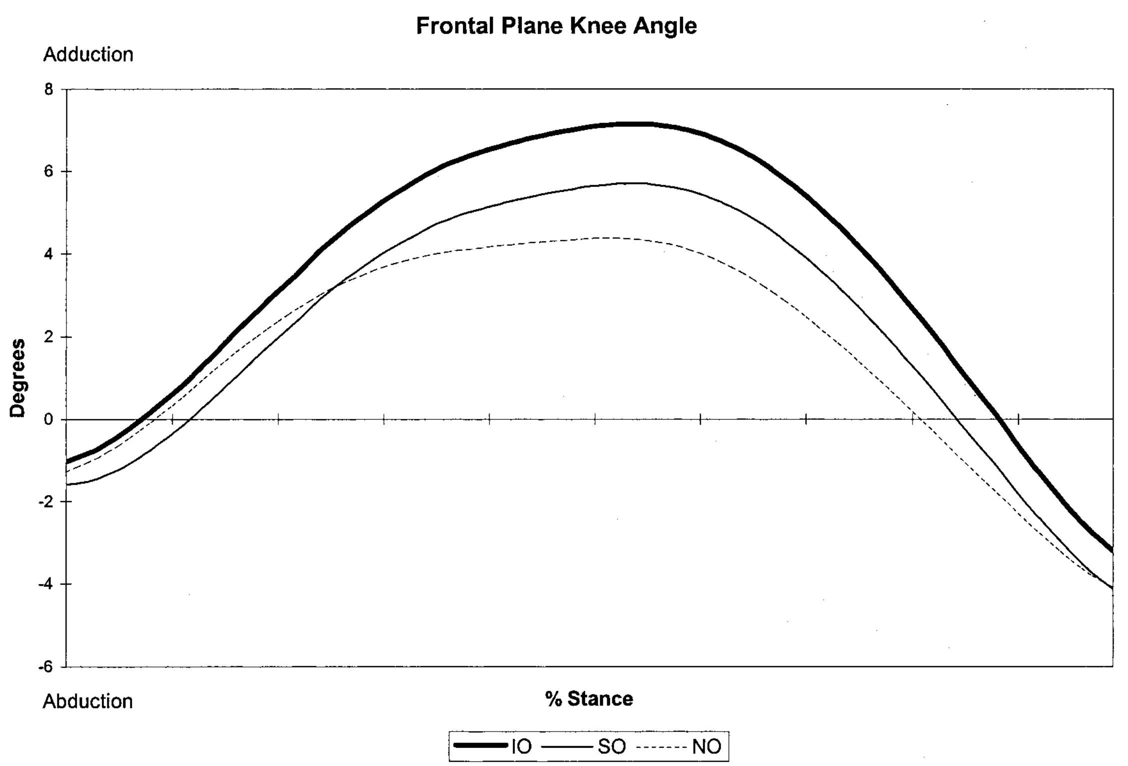

FIGURE 6—Group means for knee angle for abduction and adduction. Note the progression of peak knee adduction with increased orthotic correction. Standard deviations are not presented for clarity of means.

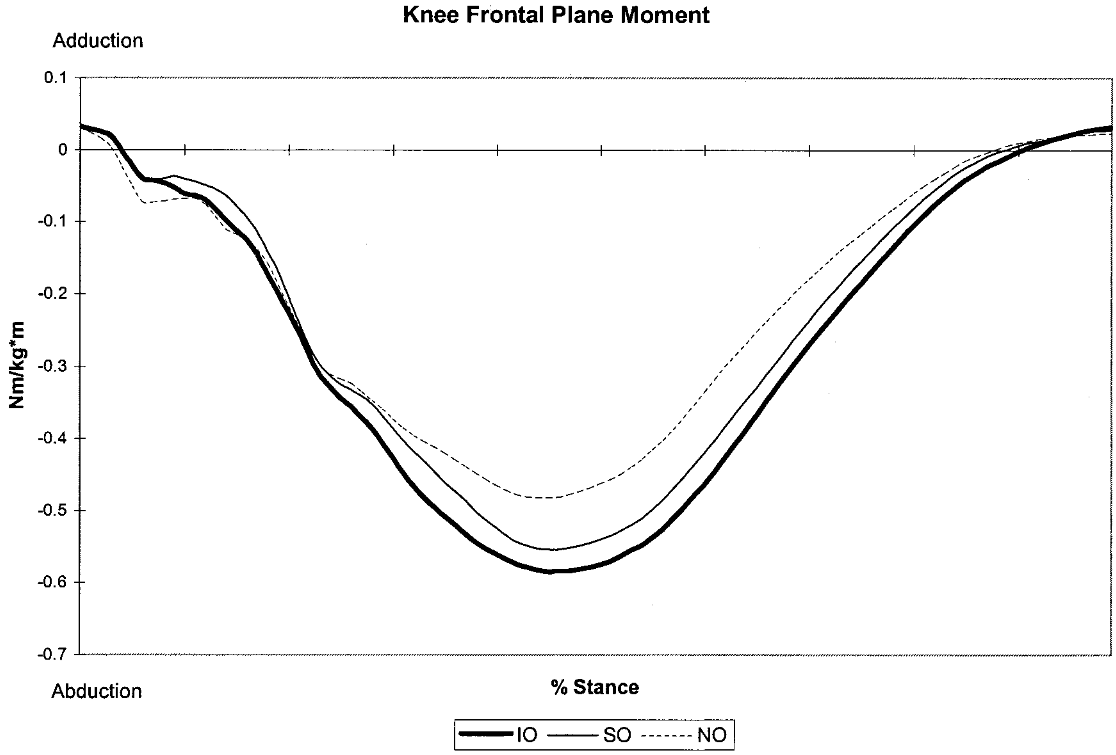

FIGURE 7—Group means for knee moment for abduction and adduction. Standard deviations are not presented for clarity of means.

is possible that these individuals felt unstable running without their orthoses and actively held their rearfoot in inversion in the no orthoses condition causing the inverted orthoses condition to be more everted. Conversely, subjects may have increased their rearfoot eversion during the inverted orthoses condition to compensate for the increased midfoot inversion provided by the shell of the device. The calcaneus is contained by a shallow heel cup and therefore has some freedom of movement in an orthotic device. Therefore, it is possible that the heel was also sliding within the shoe on the orthotic device. However, the midfoot is much contained by the rigidity of the shell of the orthoses. Therefore, orthotic devices may provide more control of the midfoot than the rearfoot. It is likely that evaluation of the midfoot may provide more complete information regarding the exact control of orthotic devices. This has yet to be established due to the difficulty in measuring midfoot motion with current motion analysis techniques.

Although mean rearfoot kinematics were unchanged in the orthotic conditions, rearfoot kinetics demonstrated significant differences. There was a systematic reduction across the conditions in both rearfoot inversion moment and negative work. This is possible, as orthoses have been shown to increase force in the midfoot in individuals wearing custom-molded orthoses (29). Although the rearfoot had a variable response in position, the orthoses in the current study likely progressively provided greater vertical force to the midfoot, thereby decreasing the force generation necessary by the muscles responsible for controlling eversion. The muscles at the ankle contribute significantly to the net moment about that joint. A decrease in both of these variables implies a reduction in the load of the muscles that control eversion (i.e., posterior tibialis). Although these runners experienced a variety of running injuries, all of them have been associated with excessive pronation, a movement controlled by the primary inverters of the rearfoot. These results partially suggest that the inverted orthotic reduced the work done by the rearfoot inverters to a greater degree than that of the standard device. These findings provide a possible explanation for the resolution of symptoms that was associated with the inverted device (and not in the standard orthoses). The inverted orthotic may have provided enough external stability in the frontal plane that the demand on the structures controlling eversion was reduced and the symptoms relieved. Future studies should examine the EMG of the rearfoot inverters to further validate whether the activity of the inverters decreases with inverted orthotic use.

Although not statistically significant, peak power absorption decreased by 23% and 34.6% between the no orthoses and standard conditions and between the no orthoses and inverted orthoses conditions, respectively (Table 2). This lack of significance may be due to a low number of subjects included in the study. A *post hoc* analysis suggests that greater than 40 subjects would have been necessary to have adequate statistical power for these variables. An important limitation of this study was that there was only one clinician referring subjects to the Motion Analysis lab for this study and the inverted orthotic technique is only used for severe patients. This made recruitment of a large number of subjects difficult.

Although there was no change in the frontal plane motion of the rearfoot, there was a progressive increase in knee adduction and knee abduction moment between the no, standard, and inverted orthoses conditions. Moving the knee away from a valgus alignment is thought to be beneficial as this position takes stress off the medial structures at the knee and reduces compression in the lateral compartment. However, the associated increase in the knee abduction moment may also have detrimental effects. It is possible that an increased joint moment place greater demands on the lateral structures of the knee. This increased load could increase the risk of overuse of the lateral structures of the knee such as iliotibial band. None of the subjects in the current study developed lateral knee pain. However, larger epidemiologic studies, including those with orthotic devices inverted beyond 25°, are needed to explore whether these devices place the wearer of such extreme devices at greater risk for lateral knee injuries.

Knee internal rotation was unchanged in both orthotic conditions. However, tibial internal rotation (tibia with respect to the foot) was progressively increased between the no, standard, and inverted conditions. This was surprising, as rearfoot motion did not change with the orthoses. Therefore, one would not have expected a change in tibial rotation due to its coupled motion with the calcaneus. The tibia can also be influenced by midfoot motion through its articulation with the talus and subsequent link to the talonavicular joint. However, the rigidity of the shell of the device likely decreased the amount of talonavicular motion in the orthotic conditions. This would result in an expected associated decrease in tibial internal rotation, opposite to what was found. In light of this increased tibial internal rotation, the lack of change in knee rotation was also surprising. For the tibial internal rotation to increase with no increase in knee rotation, femoral internal rotation must have also increased in the standard orthoses and inverted orthoses conditions. It is possible that this served as a compensatory strategy in response to the orthoses, although not measured in the current study. Additionally, foot type was not characterized in the current study. Foot type has been previously related to changes in coupling between the rearfoot and tibia (22,36).

In conclusion, these data suggest that inverted orthoses significantly decrease the inversion moment and work at the rearfoot and increase knee adduction and abduction moment when compared with standard and no orthoses conditions. Decreasing the eccentric demand on the primary inverters decreases their risk for overuse and possible injury and may explain, in part, the resolution in symptoms experienced with the use of the inverted orthoses. However, the use of the inverted orthoses may increase the load on the lateral structures of the knee, placing that individual at increased risk for knee injury. Further studies of pathology-specific populations will help elucidate the mechanisms behind the inverted orthoses' success, providing clinicians additional evidence for deciding the most appropriate patients for these devices.

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