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Influence of a custom foot orthotic intervention on lower extremity dynamics in healthy runners

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Abstract

Objective. To investigate the influence of a custom foot orthotic intervention on the lower extremity dynamics in healthy runners.

Design. Three-dimensional kinematics and kinetics were collected on 15 female runners (>15 miles per week) while each performed the over-ground running trials in either a shoe only or a shoe + custom foot orthotic condition. Kinematic and kinetic variables were analyzed using Paired Sample *t*-tests.

Background. Custom foot orthotics are frequently prescribed treatment modality for the management of overuse running injuries. Although it is generally accepted that a custom foot orthotic intervention produces positive clinical outcomes, it remains unclear what influence this therapeutic modality has on the dynamics of the lower extremity.

Methods. Each subject performed five acceptable over-ground running trials (3.6 m s^{-1} ±5%) with and without the custom foot orthotic intervention in a running shoe. Selected maximum ankle and knee joint angles and moments were measured during the stance phase.

Results. While wearing the custom foot orthotic, subjects exhibited significantly decreased maximum values in rearfoot eversion angle, rearfoot eversion velocity and internal ankle inversion moment.

Conclusions. In this sample of healthy female runners, the custom foot orthotic intervention led to significant decreases in maximum values for ankle dynamics in the frontal plane and in the sagittal plane of the knee joint.

Relevance

It remains unclear how a custom foot orthotic intervention influences lower extremity dynamics to produce positive clinical outcomes. Furthering our understanding of the dynamic influence will not only inform improved prescription and manufacturing practices but may provide insight into the mechanisms that cause overuse injuries.

Keywords: Custom foot orthoses; Overuse running injuries; Kinematics; Kinetics

1. Introduction

It has been estimated that approximately 40 million North Americans are recreational runners (McKenzie et al., 1985). Runners continue to be among the most commonly injured athletes (Baquie and Brukner, 1997). Hreljac et al. (2000) estimated that 27–70% of all runners would sustain an injury in any given year. In terms of injury site, the knee continues to be the primary location with 43–48% of all injuries occurring at the knee (Clement et al., 1981; McIntyre et al., 1991; Taunton et al., 2002).

Custom foot orthoses (CFOs) are frequently prescribed in the conservative management of overuse running injuries (Root, 1994; Williams et al., 2003; Mundermann et al., 2003). While it is generally accepted that this therapeutic modality produces positive clinical outcomes (D'Ambrosia, 1985; Saxena and Haddad, 2003), it remains unclear how this intervention influences the dynamics of the lower

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extremity during running. Prior research on the dynamic influence of CFO intervention during running has primarily focused on rearfoot (Bates et al., 1979; Smith et al., 1986; Novick and Kelley, 1990; McCulloch et al., 1993) and tibial kinematics (Nawoczenski et al., 1995). More recent studies have investigated the effects on both lower extremity kinematics and kinetics (Mundermann et al., 2003; Butler et al., 2003; Laughton et al., 2003; Williams et al., 2003).

Several investigations have employed CFOs in running studies (Bates et al., 1979; Smith et al., 1986; Baitch et al., 1991; Nawoczenski et al., 1995; Butler et al., 2003; Laughton et al., 2003; Mundermann et al., 2003; Williams et al., 2003). These studies have reported significant decreases in: (1) maximum rearfoot eversion angle (Bates et al., 1979; Smith et al., 1986), (2) maximum rearfoot eversion velocity (Smith et al., 1986), (3) maximum internal ankle inversion moment (Mundermann et al., 2003; Williams et al., 2003), (4) impact peak and maximum vertical loading rate (Mundermann et al., 2003), and (5) maximum tibial internal rotation angle (Nawoczenski et al., 1995).

Results from foot orthotic studies have been generally equivocal. The variability in results may be partly attributed to intervention design and subject inclusion criteria. For example, there have been several interventions that have been termed foot orthotics. These include ethyl vinyl acetate (EVA) wedges, modified manufacturer insoles (Stacoff et al., 2000) or posted inserts (Eng and Pierrynowski, 1993). CFOs are manufactured from a three-dimensional volumetric impression of the foot by an accredited laboratory and are typically prescribed by podiatrists, physical therapists and sports medicine physicians (Root, 1994). In podiatric practice, orthoses are prescribed to address the specific needs of the patient. Several research studies have dispensed identical designs for each subject in order to limit the confounding effects of the intervention (Mundermann et al., 2003). For example, the functional orthotic shell material and degree of intrinsic and extrinsic posting are identical for each of the devices dispensed. It could be argued that employing identical orthotic designs for all subjects could be equally or more of a confounding given the resulting device may not be comfortable or suitable for the subject needs.

Early studies by Bates et al. (1979) and Smith et al. (1986) included subject-specific designs but employed two-dimensional kinematic analyses. It has been shown that 2D analysis becomes inaccurate as the lower extremity rotates in the transverse plane (McClay and Manal, 1999). Nawoczenski et al. (1995) employed a 3D kinematic analysis and dispensed orthoses that were designed for the individual needs of the subject but the analysis was limited to the frontal plane motion and transverse motion of the tibia. This study did not include any analysis of the knee or kinetic analysis.

Another interesting issue is whether the investigation included subjects who are: (1) healthy and otherwise not candidates for this therapeutic modality or, (2) individuals with a history of running injury who typically are candidates. Many studies have included subjects who were injury-free or who did not exhibit the dynamics typically associated with overuse injury. For example, there is a clinical notion that excessive rearfoot eversion (Hintermann and Nigg, 1998) or foot architecture (Nigg et al., 1993; Williams et al., 2001) may be associated with overuse running injuries. Although a relationship has yet to be established (Messier and Pittala, 1988; Hreljac et al., 2000), CFO research has not always included subjects who would typically be clinical candidates for the intervention. We recognize the need to investigate healthy subjects and to create an understanding of how the intervention influences dynamics that can then be used to study effects in a clinical sample of runners. By studying healthy subjects, pain can be controlled as a confounder and the results may help researchers make inferences about how the intervention influences a clinical sample.

Prior research on healthy subjects has not employed true custom or subject-specific orthotic designs (Mundermann et al., 2003) or a thorough analysis of 3D kinematics and kinetics when subject-specific designs were incorporated (Nawoczenski et al., 1995). To date, there are no published studies that incorporate subject-specific CFO designs where a thorough 3D kinematic and kinetic analysis has been performed on healthy subjects. In addition, most research has only reported on variable maxima and no study has analyzed kinematic and kinetic variables at discrete points throughout the stance phase. For example, maximum vertical loading rate and impact peak occur immediately following heel strike at approximately 10–15% of the stance phase. Therefore, statistical analysis of the rearfoot angle, rearfoot velocity and internal ankle inversion moment during discrete points (i.e. 5%, 10%, 15%, 20% of stance) of this time frame may provide insight on how CFO intervention influences these variables during the time frame these events occur. Prior research has only focused on the rearfoot and maximum values of various dynamic variables and, although important, may not reveal the entire story. It is plausible that the explanation for the success of this intervention is far more complex and may lie in the kinetics of lower extremity dynamics. Intuitively, it is also reasonable to speculate that the influence that leads to reported positive clinical outcomes might occur in the initial 50–60% of stance. Given that the distal end of the typical orthotic functional shell extends just proximal to the metatarsophalangeal joint line, it is possible that dynamic influences are realized from foot contact to approximately midstance.

Therefore, the purpose of this study was to determine the short-term influence of a CFO intervention on the 3D kinematics and kinetics of the ankle and knee throughout the entire stance phase in a group of healthy female runners who exhibited a maximum rearfoot eversion angle $\leqslant 10°$ during running. It was hypothesized that the short-term influence of the CFO intervention would not lead to significant changes in lower extremity kinematics or kinetics of healthy runners exhibiting normal limits of maximum rearfoot eversion.

2. Methods

2.1. Subjects and materials

Fifteen healthy female runners were included in the study. Their average height, mass and age were 1.62 m, 60.0 kg, and 21.3 years, respectively. All subjects were recreational and competitive runners (15+ miles per week), injury-free in the six months leading up to the study and had not worn custom foot orthoses in the past. Each subject signed an informed consent in accordance with the university regulations. An a priori sample size prediction was performed using SAS v.8.2 and data from the literature (Bates et al., 1979; McClay and Manal, 1999). A sample size of 15 was estimated for a minimal statistical power of 0.80 and $\alpha = 0.05$.

Custom foot orthoses were designed and manufactured to suit the needs of the individual subject. Orthoses were fabricated from a neutral suspension cast and manufactured by an accredited laboratory (Paris Orthotics Ltd., Vancouver, BC, Canada). Orthoses were cast corrected to calcaneal vertical and intrinsically posted with an additional 5° of inversion. The shell material was either 3 mm polypropylene or copolymer depending on the mass of the subject. Each device had an extrinsic rearfoot post and a full-length cushion top cover made of either EVA or Neoprene. No forefoot extrinsic posting was used in any of the designs.

Running trials with and without the CFO intervention were performed overground in a New Balance 801 running shoe. This particular running shoe has no heel counter so that a calcaneal marker set could be fastened directly onto the skin of the calcaneus of each subject. The forefoot region of the shoe, however, is sufficiently fitted such that the subjects could run comfortably without worry that the shoe would fall off.

2.2. Experimental set-up

Kinematic and kinetic data were acquired from the right lower extremity of all subjects. Three-dimensional kinematic data were collected using an eight-camera, Qualisys Pro Reflex capture system (Qualisys, Inc., Gothenberg, Sweden). Ground reaction force data were collected using an AMTI force platform (AMTI, Inc., Watertown, MA, USA). Running speed was monitored by recording the time between two photoelectric sensors placed at each end of the testing zone. Kinematic data were collected at 240 Hz while ground reaction force data were collected synchronously at 1920 Hz.

2.3. Protocol

Four reflective tracking markers were positioned on a rigid plate and positioned on the thigh and leg. A rigid triad was also positioned on the posterior calcaneus. Prior to each individual data collection, a standing calibration trial was collected with the subject in quiet stance. For the standing calibration, additional calibration markers were positioned in order to define the individual segment geometries and segment coordinate systems. The calibration markers were positioned on the skin overlying: (1) the right and left greater trochanters, (2) medial and lateral femoral condyles, (3) medial and lateral maleoli, and (4) the heads of the first and fifth metatarsals. Following the standing calibration, these markers were removed with only the tracking markers remaining. Subjects then performed the over-ground running trials with the CFO intervention (shoe + CFO condition) and in a shoe only (SHOD condition). Five acceptable running trials were recorded for each condition at 3.6 m s^{-1} (±5%).

2.4. Data analysis

Kinematic data for the stance phase of each overground running trial were digitized using QTrac software (Qualisys, Inc., Gothenberg, Sweden). A nonlinear transformation technique was employed to calculate the 3D coordinates for each marker from the sets of two-dimensional coordinates from the eight cameras. The nonlinear transformation and marker tracking were performed using the Qualisys software.

Synchronized raw kinematic and kinetic signals were exported from Qualisys in .C3D format and processed using Visual 3D software (C-Motion, Inc., Rockville, MD, USA). Raw kinematic and kinetic data were low-pass filtered using a fourth order, zero-lag Butterworth digital filter. The cut-off frequencies for the low-pass filtering of kinematic and kinetic data were 12-Hz and 50-Hz, respectively. Data were interpolated to 101 data points, with each data point representing 1% of the stance phase.

3D segment and joint angles, angular velocities and internal joint moments were calculated using the Visual 3D software (Table 1). Joint kinematic and kinetic data are reported about anatomically oriented axes. 3D segment and joint angles were calculated using an x(flexion/extension) – y(abduction/adduction) – z(axial rotation) Cardan rotation sequence (Cole et al., 1993). Joint angles and angular velocities are reported as movement of the distal segment relative to the proximal segment. Segment angles are reported relative to the laboratory coordinate system. Angular velocities are calculated in Visual 3D using finite difference algorithms.

A Newton–Euler inverse dynamics approach was employed to calculate the joint kinetics. The foot, leg and thigh segments were modeled as frusta of a right cone and the pelvis was modeled as a cylinder. Visual 3D derived the anthropometric properties including segment mass, moment of inertia and center of mass from the subject body weight and Dempster's (1959) anthropometric data. The standing calibration markers defined segment lengths.

Table 1
Kinematic and kinetic parameters of interest

Kinematic parameters
(1) Angular rotation maxima
Calcaneal eversion
Rearfoot eversion
Tibial internal rotation
Tibial adduction
Femoral internal rotation
Knee flexion
Knee adduction
(2) Angular velocity maximum
Rearfoot eversion velocity
Kinetic parameters
(3) Internal joint moment maxima
Ankle inversion moment
Knee external rotation moment
Knee abduction moment
Knee flexion moment
(4) Maximum loading rate
(5) Vertical impact peak

In addition, joint centers are defined as the midpoint between the standing calibration markers for ankle and knee. Internal joint moments about the ankle and knee were calculated and reported in the coordinate system of the proximal segment.

2.5. Statistical analysis

Statistical analyses between SHOD and CFO conditions were performed using Paired Samples *t*-tests. Significance was indicated by a criterion α level of $p < 0.05$. Mean values for rearfoot eversion angle, rearfoot eversion velocity and internal ankle inversion moment at discrete points throughout the stance phase were analyzed in a similar manner. In order to further evaluate mean differences, effect size (ES) was calculated to express differences relative to the pooled standard deviation. ES was calculated as the mean difference between conditions divided by the pooled standard deviation (Cohen, 1988). Cohen (1988) proposed that ES values of 0.2 represent small differences; 0.5, moderate differences; and 0.8+, large differences.

3. Results

3.1. Kinematics

Subjects exhibited significant decreases in maximum rearfoot eversion angle (ES = 0.32, $p = 0.025$), calcaneal eversion angle (ES = 0.25, $p = 0.02$) and rearfoot eversion velocity (ES = 0.95, $p = 0.008$) while wearing the custom foot orthoses. In addition, the healthy subjects exhibited a significantly less maximum ankle dorsiflexion angle while wearing the CFO intervention ($p = 0.004$). There were no significant differences between conditions for any of the knee kinematic variables ($p < 0.05$). Further analysis of the rearfoot angle revealed that there the eversion angle

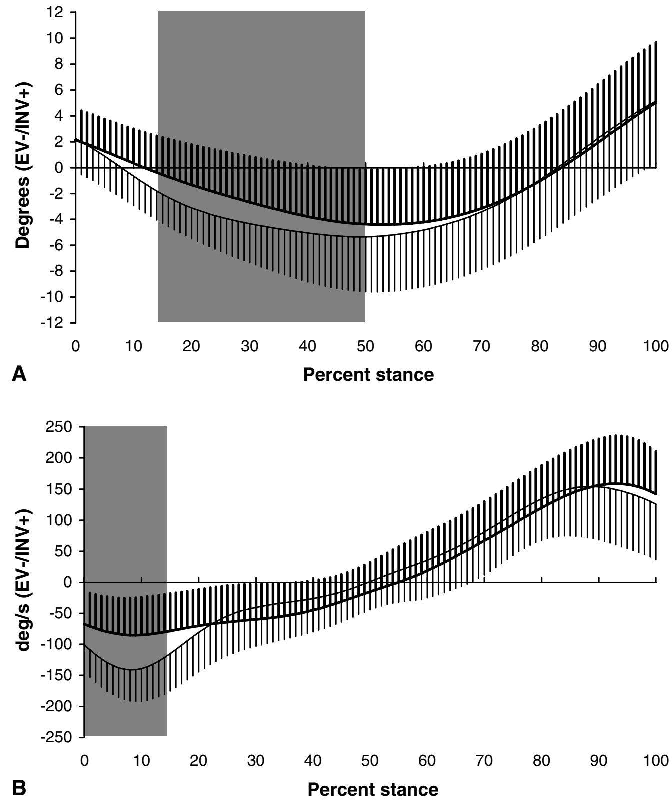


Fig. 1. (A) Ensemble averages of rearfoot frontal plane angles in degrees (EV(−)/INV(+)) during the stance phase of over-ground running. (B) Ensemble averages of rearfoot frontal plane angular velocity in degr/s (EV(−)/INV(+)) during the stance phase of over-ground running. (CFO condition (bold line) and SHOD condition (thin line). SD bars either +1 SD (bars above line) for that group or −1 SD (bars below line for that group). Grey shaded area represents early stance phase where maxima where significantly different ($p < 0.05$).

was significantly less for the CFO condition from 15% to 50% of stance ($p < 0.05$). Rearfoot eversion velocity was also significantly lesser for the CFO condition immediately following heel strike, and at 10% and 15% of stance ($p < 0.05$). Ensemble profiles of the rearfoot angle and rearfoot angular velocity are presented in Fig. 1.

3.2. Kinetics

Short-term influences of the CFO intervention in the healthy group of subjects included a significant decrease in maximum internal ankle inversion moment (ES = 0.31, $p < 0.001$). In fact, the internal ankle inversion moment was significantly less for the CFO condition from 15% to 70% of stance ($p < 0.05$). At the knee, CFO intervention resulted in significant increases in maximum internal knee extension (ES = 0.24, $p = 0.04$) and flexion moments (ES = 0.45, $p = 0.03$). There were no significant changes in maxima for either vertical impact peak or loading rate between the CFO and SHOD conditions. Ensemble profiles of the ankle inversion/eversion and knee extension/flexion moment are presented in Fig. 2.

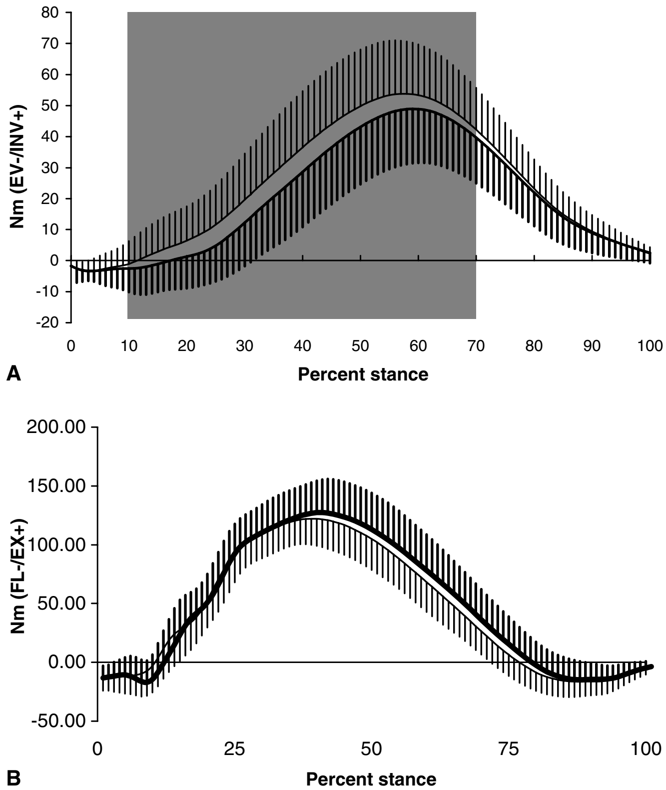


Fig. 2. (A) Ensemble average of internal ankle inversion moment in N m (EV(−)/INV(+)) during the stance phase of over-ground running. (B) Ensemble average of internal knee moment in N m (FL(−)/EX(+)) during the stance phase of over-ground running. CFO condition (bold line) and SHOD condition (thin line). SD bars either +1 SD (bars above line) for that group or −1 SD (bars below line for that group). Grey shaded area represents early stance phase where maxima where significantly different ($p < 0.05$).

4. Discussion

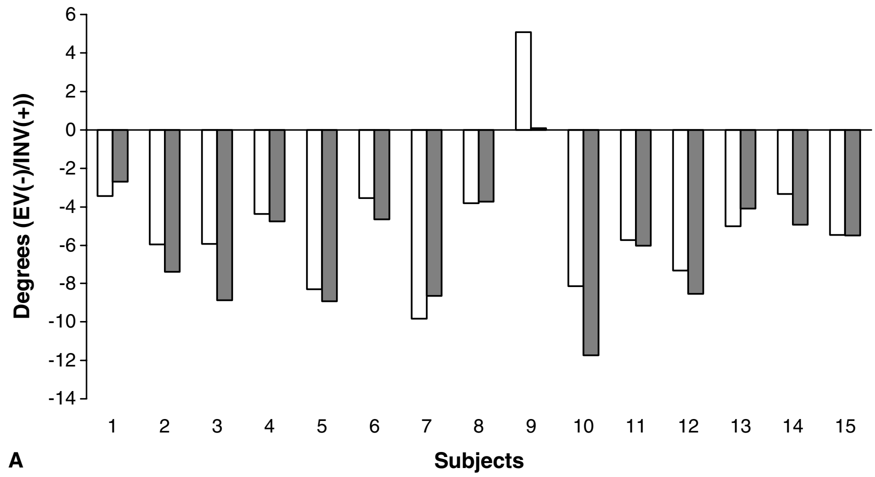
The purpose of the current study was to analyze the short-term influence of CFO intervention on ankle and knee dynamics throughout the entire stance phase in a sample of healthy runners who exhibited a within normal limits maximum rearfoot eversion angle during running. It was hypothesized that the intervention would have no influence on ankle and knee dynamics. Although this hypothesis was not completely supported, there were some interesting results that may help to explain how custom foot orthoses influence lower extremity dynamics.

Results from this investigation suggest that this short-term intervention primarily affected the rearfoot kinematics and kinetics. Similar to Bates et al. (1979) and Smith et al. (1986), the maximum rearfoot eversion angle was significantly reduced. However, it is questionable how functionally significant a statistically significant reduction of approximately 1.0° of rearfoot eversion may be (CFO condition 5.20° vs. SHOD condition 6.28°). This notion is supported by an effect size (0.32) that is considered small and thus the difference between condition means is relatively small (Cohen, 1988). Inconsistency in the results reported by previous studies for this variable may be attributed to the small magnitude of change itself or the natural variability inherent of this movement pattern. Nevertheless, maximum rearfoot eversion angle was reduced in 10 of the 15 subjects included in this study (Fig. 3A). Further statistical analysis of rearfoot eversion angle at discrete points throughout the stance phase revealed that eversion was reduced from 15% to 50% of the stance phase (Fig. 1A). Results also revealed that changes in rearfoot eversion angle were attributed to reductions in calcaneal eversion rather than tibial adduction. In the frontal plane, these data suggest that the calcaneal angle was influenced by CFO intervention whereas the leg angle went unchanged.

In addition, there was a significant reduction in maximum rearfoot eversion velocity of approximately 29 degr s^{-1} that resulted in a very large effect size (0.95) and so may be considered clinically meaningful. This result is also in agreement with the results published by Smith et al. (1986) but not supported by Mundermann et al. (2003). This finding may be important as increases in maximum rearfoot eversion velocity have been associated with overuse running injuries (Messier and Pittala, 1988; Hreljac et al., 2000). Maximum rearfoot eversion velocity was reduced in 14 of the 15 subjects included in this study (Fig. 3B). In addition, the angular velocity of rearfoot eversion was also significantly reduced from heel strike to 15% of the stance phase (Fig. 1B).

Maximum ankle dorsiflexion angle was also significantly less when the subjects wore the CFO intervention. This may in part be due to a slight heel lift that is intrinsic to the device or a reduced need for ankle dorsiflexion while the heel is supported in this manner (Bates et al., 1979). At the knee, none of the kinematic variables were influenced by the CFO intervention. This is a different result than that reported by Williams et al. (2003) who reported significant increases in maximum knee adduction angle with CFO intervention in subjects with a history of injury. There were no significant changes in leg segment, thigh segment or knee joint internal or external rotation as reported by Nawoczenski et al. (1995) and Mundermann et al. (2003).

Analysis of the internal resultant joint moments may help researchers to make inferences about the stress placed on biological tissues that produce moments about joints. In previous investigations, authors have reported on reductions in the internal ankle inversion moment with CFO intervention (Mundermann et al., 2003; Williams et al., 2003). Results of the current investigation also indicated that the maximum ankle inversion moment was significantly reduced with CFO intervention. This kinetic variable was reduced in 14 of the 15 subjects analyzed in the current study (Fig. 4A). In addition, an analysis of the entire stance phase revealed that this variable was significantly reduced from 10% to 70% of the stance phase (Fig. 2A). The posterior tibialis muscle is regarded as the major contributor to the production of internal ankle inversion moment and may act eccentrically to limit pronation (O'Connor and Hamill,



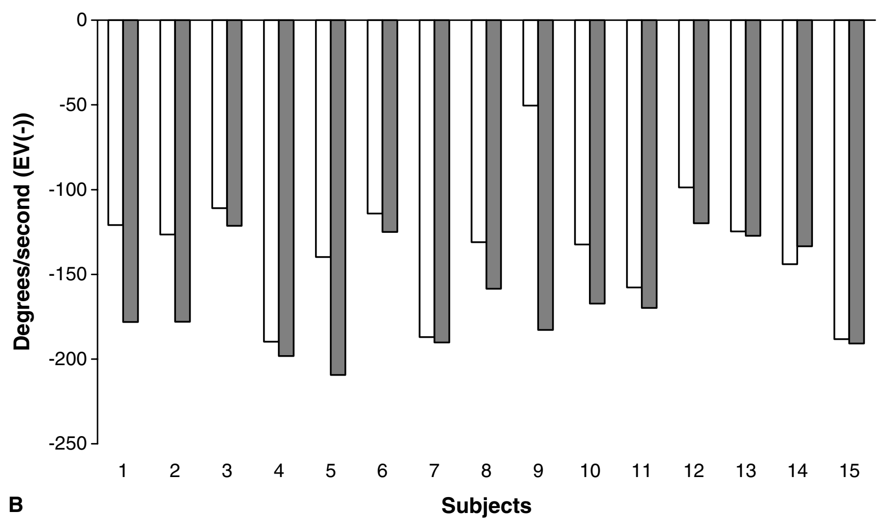


Fig. 3. (A) Individual maximum rearfoot eversion angles in degrees (EV(−)/INV(+)) during the stance phase of over-ground running. (B) Individual maximum rearfoot eversion velocities in degr/s (EV(−)) during the stance phase of over-ground running (white bars represent CFO condition; grey bars represent SHOD condition).

2004). Results of the current investigation suggest that a CFO intervention assisted in reducing the internal ankle inversion moment and thus, possibly assisted the posterior tibialis muscle in controlling subtalar joint pronation. It has been suggested that subtalar joint pronation is an essential protective mechanism that facilitates the attenuation of impact forces during running (O'Connor and Hamill, 2004; Hreljac, 2004). Hreljac (2004) has suggested that it is only when kinematic and kinetic variables extend beyond healthy physiological limits that they may be deleterious.

At the knee, this sample of runners exhibited significant increases in maximum internal knee extension and flexion moments while wearing the custom foot orthoses (Fig. 2B). Increases in the maximum knee extension moment were exhibited in 10 of the 15 subjects (Fig. 4C). Although increases in this variable were unsystematic, it is possible that increasing the knee extension moment could be deleterious to patellofemoral dynamics. The maximum knee flexion moment was also significantly increased when subjects ran in the foot orthoses. 11 of the 15 subjects exhibited greater knee flexion moment for the CFO condition (Fig. 4).

In the frontal and transverse plane, there were no significant changes in internal knee joint moments. This is in contrast to Williams et al. (2003) who reported significant increases in maximum internal knee abduction moment and significant increases in maximum knee adduction angle in a group of runners with a history of running injury. Intuitively, if the knee was in an adducted position prior to heel strike, increasing the degree of knee adduction and internal abduction moment may be deleterious to the medial compartment and lateral structures of the knee. Notwithstanding, if the knee is in an abducted attitude prior to heel strike, this influence may help position the knee in a healthier manner for those who suffer from patellofemoral problems (Powers, 2003). It is plausible that in runners with a history of injury or who are susceptible to injury, the custom foot orthoses influence the knee in a manner that helps to address deleterious knee dynamics in the frontal plane not seen in a healthy sample.

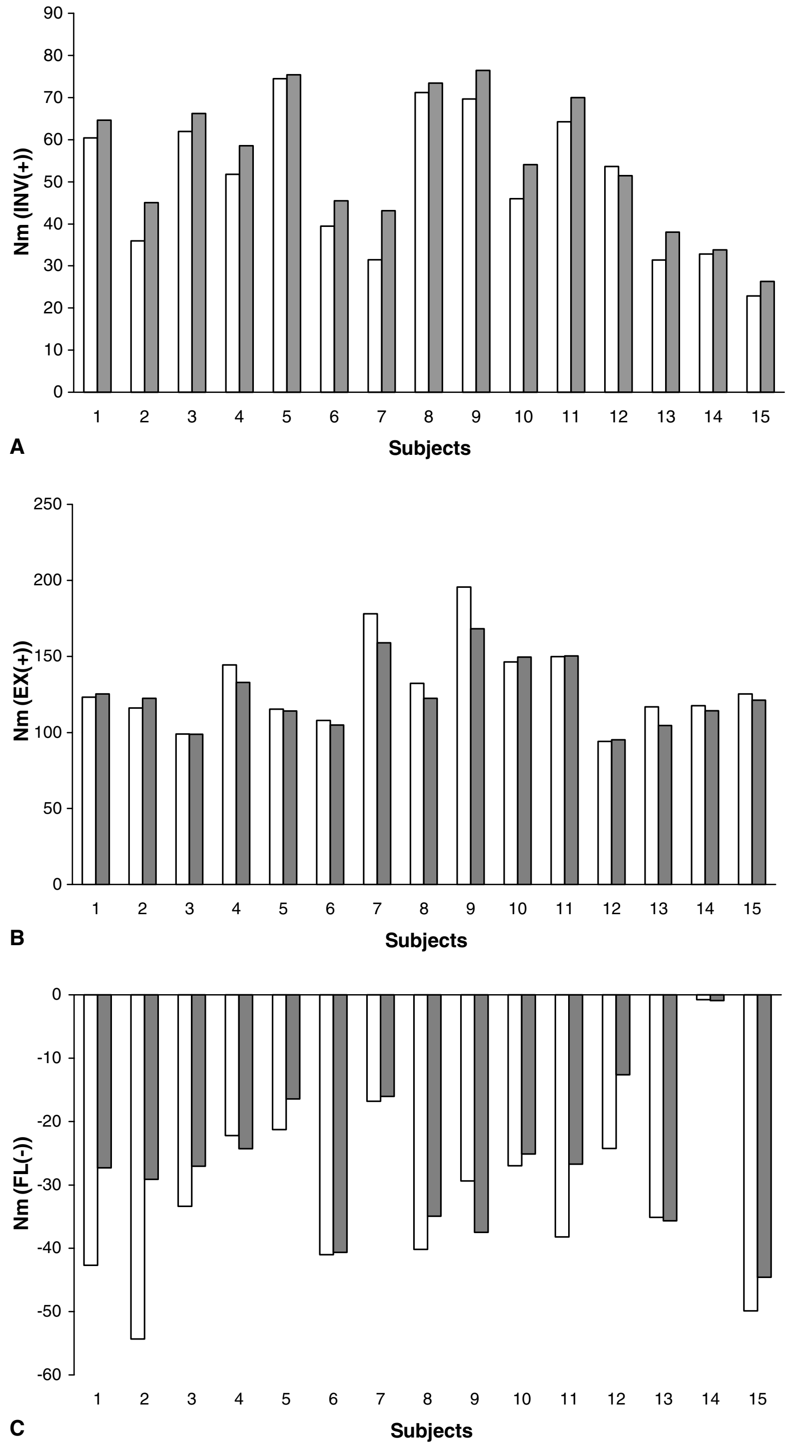


Fig. 4. (A) Individual maximum internal ankle inversion moments in Newton meters (N m) (INV +) during the stance phase of over-ground running. (B) Individual maximum internal knee extension moments in Newton meters (N m) (EX +) during the stance phase of over-ground running. (C) Individual maximum internal knee flexion moments in Newton meters (N m) (FL −) during the stance phase of over-ground running. White bars represent CFO condition; grey bars represent SHOD condition.

It is interesting that the influence of the CFO intervention was primarily realized in the initial 50–60% of the stance phase. There are several dynamic factors that occur in the initial stages of the stance phase including maximum impact peak, vertical loading rate and rearfoot eversion velocity. Mundermann et al. (2003) reported that a CFO intervention significantly reduced both the impact peak and vertical loading rate in a group of subjects who exhibited >13° of maximum rearfoot eversion. If these factors and other rearfoot kinematic factors extend beyond healthy limits, it is plausible that the positive clinical outcomes produced by CFO intervention are realized in the initial stages of the stance phase. In the current study, changes in dynamic movement patterns primarily occurred in advance of 60–70% of the stance phase.

5. Conclusions

Results from the current study reveal that short-term CFO intervention in a sample of healthy runners led to significant reductions in maximum rearfoot eversion angle and velocity, and internal ankle inversion moment. It was interesting to note that changes in lower extremity dynamics occurred primarily in the initial stages of the stance phase. Reported changes in these movement patterns in healthy subjects may help to provide insight into how CFO intervention produces positive clinical outcomes.

At the knee, kinematic variables were not significantly influenced but subjects did exhibit increases in internal knee joint moments in the sagittal plane. Further research is required to explore the effect of this intervention in a clinical sample as well as subjects who exhibit potentially deleterious dynamic patterns.

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