

# Effect of foot orthotics on rearfoot and tibia joint coupling patterns and variability

Reed Ferber<sup>a,\*</sup>, Irene McClay Davis<sup>b,c</sup>, Dorsey S. Williams III<sup>d</sup>

<sup>a</sup> Human Performance Laboratory, Faculty of Kinesiology, 2500 University Drive NW, University of Calgary, Calgary, Alta., Canada T2N 1N4

<sup>b</sup> Department of Physical Therapy, University of Delaware, Newark, DE 19716, USA

<sup>c</sup> Joyner Sportsmedicine Institute, Mechanicsburg, PA 17111, USA

<sup>d</sup> Department of Physical Therapy, East Carolina University, Greenville, NC 27858, USA

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## Abstract

The purpose of this study was to compare joint coupling patterns and variability of the rearfoot and tibia during running in subjects who were treated with two types of orthotic devices to that of controls. Eleven subjects with various lower extremity injuries were treated unsuccessfully with a standard orthotic, and then successfully with an inverted orthotic. Three-dimensional kinematic data were collected while subjects ran without orthoses and then in standard and inverted orthoses. Eleven healthy subjects ran without orthoses for comparison. The rearfoot inversion/eversion and tibial internal/external rotation joint coupling pattern and variability relationship was assessed using a vector coding technique. It was hypothesized that when the treated runners ran without orthotic devices, they would exhibit lower joint coupling angles and lower joint coupling variability compared to the controls. In addition, it was hypothesized that there would be no difference in the coupling angle or coupling variability between the standard and no orthotic conditions of the treated runners. Finally, it was hypothesized that coupling angle would decrease and variability would increase in the inverted versus the standard and non-orthotic conditions. No significant differences in joint coupling pattern or variability were observed between the treated and control subjects. In addition, no significant differences were noted between the orthotic conditions in the treated group. These results suggest that foot orthotic devices do not produce significant changes in rearfoot–tibial coupling. Therefore, the relief experienced with the inverted orthotic is likely due to factors other than alterations in this coupling.

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## 1. Introduction

Foot orthotic devices have been shown to be extremely efficacious for running related injuries (Eggold, 1981; Kilmartin and Wallace, 1994). However, the mechanism responsible for their success is not very well understood. While results have been mixed, most studies show that orthotics do reduce some aspect of rearfoot motion (Bates et al., 1979; Johanson et al., 1994; Rodgers and Leveau, 1982).

Blake (1986) proposed the inverted orthosis for patients with injuries related to pronatory problems, who had not responded to standard orthotic intervention. According to Blake (1986), this device provides significantly greater support in the arch compared to the standard orthosis. Baitch et al. (1991) later reported that the inverted orthosis resulted in significantly reduced rearfoot motion than a standard device. However, he did not extend his assessment to the knee.

The majority of running injuries are located at the knee (Taunton et al., 2002). Foot orthotic devices have been shown to be very effective in reducing knee pain (Eng and Pierrynowski, 1993). Alteration of rearfoot motion is likely to influence knee motion due to the coupling that has been shown to occur in the lower

\*Corresponding author. Tel.: +1-403-220-7004; fax: +1-403-284-3553.

E-mail address: rferber@kin.ucalgary.ca (R. Ferber).

extremity (Lundberg et al., 1989; Inman, 1976). During the first half of the stance phase, the calcaneus everts and the head of the talus internally rotates (Lundberg et al., 1989; Inman, 1976). The tibia internally rotates with the talus due to the tight articulation of the ankle joint mortise. After reaching a peak in midstance, these motions reverse such that inversion and tibial external rotation continue throughout the last half of stance. In this way, rearfoot motion influences the motion of the knee.

One way in which investigators have studied the coupling of eversion (EV) and tibial internal rotation (TIR) is through the evaluation of the relative excursions of these motions. An EV/TIR excursion ratio is formed by dividing the excursion of eversion by that of tibial internal rotation, over the time period from heel strike to the respective peak values (occurring around midstance). As there is normally more eversion than tibial internal rotation, this ratio has been reported to vary between 1.0 and 1.8 (McClay and Manal, 1997; Nawoczenski et al., 1998; Stacoff et al., 2000; Williams et al., 2001).

It has been shown that runners with low arches (thought to be related to excessive eversion) exhibited higher EV/TIR excursion ratios (Nawoczenski et al., 1998; Stacoff et al., 2000; Williams et al., 2001). As foot orthotics are typically designed to control rearfoot eversion, they will likely reduce the relative amount of eversion to tibial internal rotation motion and thus alter their joint coupling relationship. However, Nawoczenski et al. (1995) evaluated the effect of standard orthoses on the EV/TIR ratio of healthy runners and found an increase in the EV/TIR excursion ratio mainly due to reduced TIR.

The EV/TIR excursion ratio is a single value which represents the average joint coupling over the first half of stance. Heiderscheit et al. (2002) proposed using a vector coding method (Sparrow et al., 1987) to provide a measure of continuous joint coupling. In this method, the relative excursions between successive sampled data points of an angle–angle diagram are calculated, and the resultant angle (referenced to the horizontal) is computed between these points. This process is then repeated across the stance period. In addition to the coupling angle, Heiderscheit et al. (2002) also recommend evaluating the within subject variability of that angle as a way of assessing injury risk. These authors reported that runners with patellofemoral pain syndrome exhibited no differences in joint coupling patterns, but lower variability than the healthy controls. Their findings support the premise of others that variability in movement is necessary to adapt to a changing environment and reduce the risk for overuse injuries (Holt et al., 1995; Hamill et al., 1999).

The influence of foot orthoses on lower extremity joint coupling pattern or variability has not been

investigated in a patient population. The vector coding approach may provide new insight into injury mechanisms as well as the mechanism behind the success of orthotic intervention. Therefore, the purpose of this study was to compare joint coupling patterns and variability of the rearfoot and tibia during the stance phase of running in subjects who were treated with two types of orthotic devices to that of healthy controls. In addition, joint coupling patterns and variability were compared in these treated subjects when they ran without orthotics (NO), in a standard orthotic device (STD) that did not provide relief of their symptoms and an inverted device (INV) that did provide relief. Based on angle–angle plots derived from the vector coding technique, it was hypothesized that treated runners, when running without orthoses (NO), would exhibit reduced coupling angles (due to reduced rearfoot eversion relative to tibial internal rotation) and reduced coupling variability than the control group. It was also hypothesized that there would be no significant difference in coupling angle or variability when treated subjects ran in the STD device compared to the NO device condition. Finally, it was expected that there would be a significant increase in the coupling angle and an increase in the coupling variability when treated subjects ran in the INV device compared to the NO and STD conditions.

## 2. Methods

Based on a priori power analyses ( $\beta = 0.20$ ;  $P = 0.05$ , effect size and variability from previous literature: Baitch et al., 1991), 11 runners (5 males, 6 females) who had been treated with foot orthoses by a single physical therapist were recruited for this study. The mean age, body mass, and body height of subjects were 29.9 yr ( $\pm 12.2$  yr), 69.2 kg ( $\pm 15.7$  kg), and 169.8 cm ( $\pm 12.9$  cm), respectively. The runners presented with a variety of running related injuries (Table 1) and a licensed physical therapist performed a standard clinical examination and visually noted that each subject exhibited excessive static rearfoot or mid-foot pronation. These subjects were initially fitted with a standard foot orthosis and each STD orthosis was made with a graphite shell and posted in 4° of rearfoot varus and intrinsic forefoot posts were balanced from non-weight bearing plaster casts. All of these subjects had persistence of symptoms for 1–2 months when using the STD orthotic. Each subject was then fitted with INV graphite orthoses posted between 15° and 25° depending upon the severity of their structure and symptoms. Orthoses were individually prescribed for each limb. All of these 11 subjects experienced relief of their symptoms with the INV orthosis and had been wearing these devices a minimum of 4 months prior to the data collection. All

Table 1  
Subject characteristics, degree of INV orthoses angle, and injury diagnoses

Subject	Gender	Age (yr)	Mass (kg)	Deg. INV	Limb	Diagnosis
1	F	41	45.2	25	Left	Posterior tibial tendonitis
2	M	49	100	25	Left	Plantar fasciitis
3	M	23	64.1	15	Left	Anterior comp syndrome
4	M	23	65.5	15	Left	Anterior comp syndrome
5	M	30	94.1	25	Right	Anterior comp syndrome
6	F	18	49.2	25	Right	Anterior comp syndrome
7	F	20	68.7	25	Right	Patellofemoral pain syndrome
8	F	18	66.2	25	Right	Plantar fasciitis
9	F	18	62.1	15	Right	Plantar fasciitis
10	M	52	78.2	15	Right	Plantar fasciitis
11	F	37	68.2	15	Left	Patellofemoral pain syndrome

subjects continued to wear the devices once their symptoms resolved as they believed their problems would recur without them.

Data from 11 runners (5 males, 6 females) served as the controls (CON) and they had not sustained a lower extremity injury in at least 2 years. In addition, foot orthotic devices had never been prescribed for these runners. The mean age, body mass, and body height of subjects were 21.4 yr ( $\pm 3.1$  yr), 67.0 kg ( $\pm 11.7$  kg), and 173.1 cm ( $\pm 9.3$  cm), respectively. These subjects were randomly chosen from a normative database of runners previously collected while running without orthoses and the limb chosen for analysis was matched to that of the treated group. Prior to participation, each subject signed a consent form approved by the University's Human Subjects Compliance Committee.

Retro-reflective markers for tracking 3D movement were placed on the thigh, shank, pelvis, and rearfoot (Fig. 1). Anatomical markers defining the joint centers were placed over the following locations: bilateral greater trochanters, medial and lateral femoral condyle, medial and lateral malleoli, heads of the 1st and 5th metatarsals. A static standing calibration was collected prior to each condition. Only the markers were removed, leaving the bases on the subjects so that anatomical marker placement was consistent between conditions. Three rearfoot markers were placed directly on the heel and extended through windows cut in the shoes (Fig. 1). These windows allowed for unabated motion of the markers on the heel. It was determined that these holes result in approximately 10% decrement in heel counter stability as measured by an Instron materials testing device (Canton, MA, USA). Subjects ran along a 25 m runway at a speed of 3.65 m/s ( $\pm 5\%$ ) which was a comfortable running speed for all subjects. Running speed was monitored using photoelectric cells placed 2.86 m apart along the runway. The affected limb of the treated subjects and the matched limb of the control subjects were chosen for analysis. Eight trials were collected for each of the randomized conditions: NO, STD, and INV. The

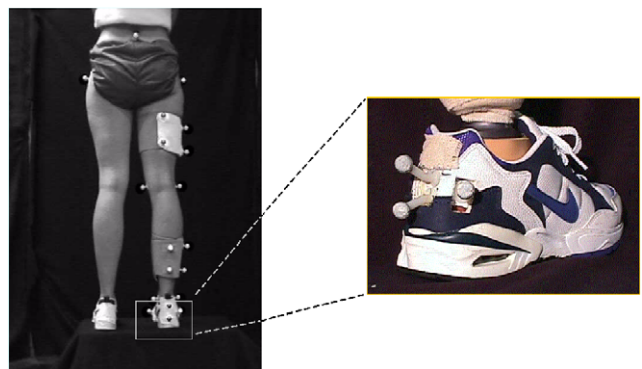


Fig. 1. Retroreflective marker placement on the tested lower extremity.

control subjects completed 8 trials for the NO condition. The first five acceptable trials were used for the analysis since five trials have been shown to provide stable mean kinematic and kinetic data (Bates et al., 1992; Diss, 2001).

Kinematic data were collected with a passive, 6-camera, 3D motion analysis system (VICON, Oxford Metrics, UK). Kinematic data were sampled at 120 Hz and low-pass filtered at 8 Hz with a fourth-order zero lag Butterworth filter. Kinetic GRF data were collected using a force plate (BERTEC Corp, Worthington, OH, USA). GRF data were synchronized with the kinematic data and were collected at 960 Hz and low-pass filtered at 50 Hz with a fourth-order zero lag Butterworth filter. Trials were normalized to 100% of stance and five trials were averaged for each subject. MOVE3D software (MOVE3D, NIH Biomotion Laboratory, Bethesda, MD, USA) was used to calculate the three-dimensional kinematic joint angles. While it is understood that the transverse plane motion is susceptible to soft tissue error, studies within this laboratory (Ferber et al., 2002) have reported good within-day and between-day variability for the transverse plane motion of the tibia with respect to the foot. In addition, comparisons among the orthotic conditions are made

within the same subjects, and thus the errors should be consistent across conditions.

Since it is recognized that the markers on the lower leg track the motion of the entire shank, this segment will be referred to as the tibia to remain consistent with notation used in the literature (Hamill et al., 1999; Heiderscheit et al., 2002; Nigg et al., 1999). Angle–angle plots of rearfoot eversion/inversion (RF) and tibial internal/external rotation (TIB) for each trial were created. The coupling angle was determined using a modification of a vector coding technique suggested by Heiderscheit et al. (2002). The absolute resultant vector between two adjacent data points during the stance phase of running was calculated (Eq. (1)). Following conversion from radians to degrees, the resulting range of values for  $\phi$ , or coupling angle, was 0–90° (Fig. 2).

$$\phi_i = \text{abs}[\tan^{-1} (y_{i+1} - y_i / x_{i+1} - x_i)]$$

where  $i = 1, 2, \text{ and } n$ . (1)

Thus, with RF motion plotted on the abscissa and TIB motion plotted on the ordinate, a coupling angle of 45° would indicate equal amounts of rearfoot frontal plane and tibial transverse plane motion. An angle greater than 45° indicates greater tibial transverse plane motion relative to rearfoot frontal plane motion (Fig. 2).

For the purpose of analyzing the RF–TIB coupling angle and variability within specific regions of stance, each relative motion plot was first normalized to 100% of stance. Each relative motion plot was then divided into 4 phases that were selected according to discrete kinetic events determined from vertical ground reaction force data. Phase 1 ranged from heel strike to initial loading (~0–20% of stance), phase 2 from initial loading to acceptance of full body weight (~20–50% of stance), phase 3 from acceptance of full body weight to half the distance to toe-off (~50–75% of stance), and phase 4 ranged from the end of phase 3 until toe-off (~75–100% of stance). To calculate the average

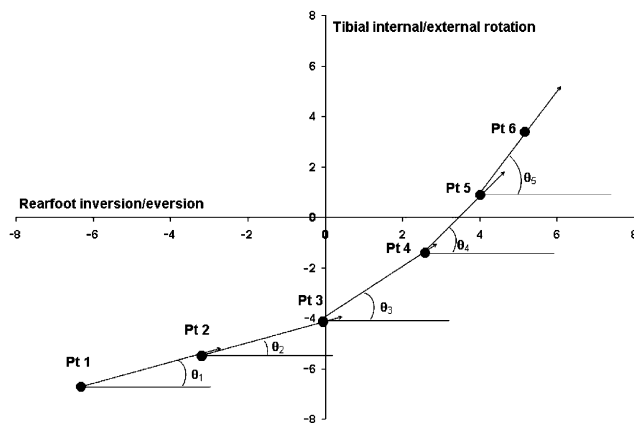


Fig. 2. Example of vector coding technique to determine joint coupling angle between the RF and TIB.

coupling angle values for each phase of stance, each data point was averaged on a point-by-point basis across the 5 trials resulting in an average trace. From the average trace, the average coupling angle for each phase of stance was calculated over time (Table 1). The standard deviation was calculated on a point-by-point basis across the 5 trials and the between-trial, within-subject joint coupling variability for each phase of stance was calculated across time for each phase of stance (Table 1).

A two-factor ANOVA (group  $\times$  phase:  $\alpha = 0.05$ ) with repeated measures on phase between the control group and each condition of the treated group were used to identify differences in RF–TIB joint coupling angle. Another two-factor ANOVA (group  $\times$  phase:  $\alpha = 0.05$ ) with repeated measures on phase between the control group and each condition of the treated group were used to identify differences in mean RF–TIB joint coupling pattern variability. Tukey post hoc tests were used to determine differences, if any, for joint coupling angle and variability for the selected hypotheses: (1) NO vs. controls, (2) NO vs. STD, and (3) INV vs. NO and STD. Hypothesis 1 was tested with a between subject design, while hypotheses 2 and 3 were tested with a within subject design.

### 3. Results

Fig. 3 shows the joint coupling pattern for the control group, as well as the treated group when running in each condition. During the first 50% of stance (phases 1 and 2) the control group exhibited relatively little change in the RF–TIB coupling pattern but this pattern fluctuated to a greater extent during the last 50% of stance (phases 3 and 4: Fig. 3). In addition, the two orthotic conditions in the treatment group were very similar to the no orthotic condition throughout stance.

The means and standard deviations for the coupling angles for each phase are presented in Table 2 and

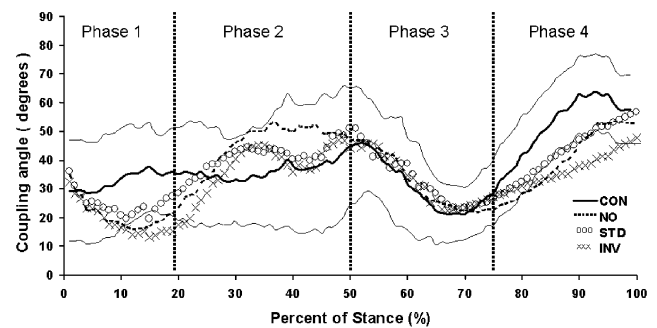


Fig. 3. Average RF–TIB joint coupling angle plot for the control group (thick and thin lines equal mean  $\pm$  1SD) and treated subjects when running in three different orthoses conditions during the stance phase of gait.



Table 2  
Summary of coupling angle and variability and *P*-values across the four phases of stance

Variable	NO <sup>b</sup>	STD	INV <sup>c</sup>	CON <sup>a</sup>
<i>Coupling angle (deg)</i>				
Phase 1	20.76 (12.40) <i>P</i> =0.98	23.71 (11.07)	19.42 (9.81) <i>P</i> =0.99	31.04 (13.51) <i>P</i> =0.28
Phase 2	40.51 (6.23) <i>P</i> =0.95	38.27 (8.79)	34.28 (7.84) <i>P</i> =0.84	35.98 (14.21) <i>P</i> =0.99
Phase 3	34.93 (8.70) <i>P</i> =0.94	34.83 (10.41)	34.14 (9.95) <i>P</i> =0.93	32.26 (10.01) <i>P</i> =0.97
Phase 4	38.24 (8.97) <i>P</i> =0.95	41.53 (10.79)	35.78 (14.98) <i>P</i> =0.99	50.25 (9.36) <i>P</i> =0.10
<i>Coupling variability (Std. dev.)</i>				
Phase 1	11.56 (6.45) <i>P</i> =0.95	12.05 (5.39)	10.88 (5.19) <i>P</i> =0.98	11.64 (5.30) <i>P</i> =0.96
Phase 2	13.40 (3.15) <i>P</i> =0.96	14.58 (3.24)	11.82 (2.67) <i>P</i> =0.65	13.48 (4.13) <i>P</i> =0.98
Phase 3	13.48 (4.28) <i>P</i> =0.99	13.81 (3.79)	13.92 (4.36) <i>P</i> =0.93	15.70 (2.43) <i>P</i> =0.94
Phase 4	8.34 (2.19) <i>P</i> =0.99	7.91 (2.04)	8.63 (2.95) <i>P</i> =0.98	10.16 (3.15) <i>P</i> =0.67

<sup>a</sup> *P*-values compared to NO condition.

<sup>b</sup> *P*-values compared to STD condition.

<sup>c</sup> *P*-values compared to STD and NO conditions.

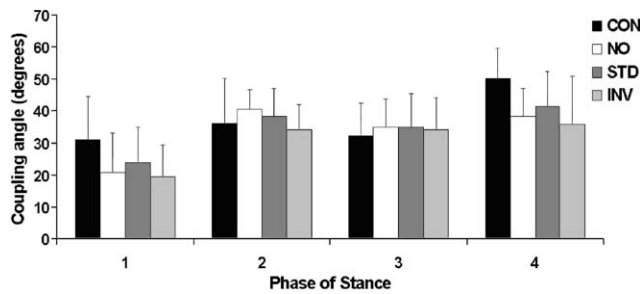


Fig. 4. Average RF-TIB joint coupling angle of the treated subjects during 3 different conditions and controls for the 4 phases of stance as determined by discrete vertical ground reaction force events.

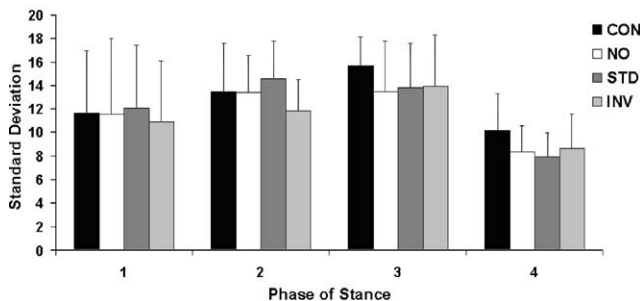


Fig. 5. Average RF-TIB joint coupling angle variability of the treated subjects during 3 different conditions and controls for the 4 phases of stance as determined by discrete vertical ground reaction force events.

**Figs. 4 and 5.** No significant differences in joint coupling angle or variability were observed between the control group and the treated group when they ran in the NO condition for any phase of stance. In addition, no significant differences in joint coupling angle or variability were observed for the treated group when running in the STD compared to the NO condition for any phase of stance. Finally, no significant differences were observed when treated subjects ran in the INV device

compared to either the NO or STD conditions for any phase of stance.

#### 4. Discussion

The purpose of this study was to compare the joint coupling pattern and variability of the rearfoot and tibia during running in treated runners vs. healthy controls to better understand the mechanisms behind running injuries and the success of the inverted orthosis. Since the treated subjects were visually observed to be excessive pronators, it was expected that their coupling angle would be reduced when they ran in the no orthotic condition compared to the healthy controls. However, no differences were found. These findings were consistent with those of [McClay and Manal \(1997\)](#) who compared the EV/TIR ratios between runners with normal rearfoot mechanics to those of healthy, but excessive pronators. They reported similar eversion excursions during the first half of stance between the groups. However, the pronator group involved in that study exhibited greater tibial excursions resulting in a surprisingly lower EV/TIR ratio than the runners with normal rearfoot mechanics.

No differences were noted in joint coupling variability between the treated and the control group in the present study. [Heiderscheit et al. \(2002\)](#) studied patients with patellofemoral pain syndrome and reported reduced lower extremity joint coupling variability for thigh rotation and leg rotation coupling compared to the uninjured leg and the control group. These results suggest that they might have had reduced flexibility in their systems placing them at greater risk of injury. However, [Heiderscheit et al. \(2002\)](#) did not determine RF-TIB coupling patterns or variability so it is difficult to compare their results with

those of present study. The subjects in the present studied also presented with a variety of running injuries, which may have added inter-subject variability in the data decreasing the power to detect differences in some cases.

The remaining questions focused on comparisons between the orthotic conditions in the treatment group. No significant differences in joint coupling angle or variability were observed when the treated subjects ran in the STD device compared to the NO device condition. This was an expected finding as the STD device did not reduce the clinical symptoms in this group of treated runners. However, despite the reduction in symptoms afforded by the INV orthoses, no increases in either coupling angles or variability were observed when compared to the NO and STD conditions. It was anticipated that running in the INV orthotic would result in a significant increase in the coupling angle. However, this was not observed in the present study. This is in sharp contrast to the findings of Baitch et al. (1991) who reported that the inverted orthosis significantly reduced the amount of rearfoot eversion compared to an STD and NO orthotic condition. However, Baitch et al. (1991) studied a 25° INV device which may have resulted in an overall greater rearfoot eversion reduction compared to the subjects in the present study who received orthoses posted between 15° and 25° depending upon the severity of their structure and symptoms (Table 1).

There are factors which may have influenced the results of this study. The study is retrospective in nature. Therefore, the joint coupling patterns seen in the treated subjects may not have been present prior to their injury. Prospective studies are needed to provide insight into the relationship between joint coupling patterns and injury. Secondly, the treated subjects were chronic orthotic users and the NO condition may not be truly indicative of the runner's mechanics prior to orthotic treatment. Therefore, true differences involving the NO condition and the INV and STD devices may be masked. Third, it is possible that inclusion of more subjects may have resulted in statistical significance for some of the comparisons. However, a sample size estimate indicated that 11 subjects was sufficient.

However, despite these limitations and based upon the results of these data, it appears that neither RF-TIB joint coupling patterns nor variability was different between treated runners and healthy controls. In addition, these results suggest that neither the STD nor the INV orthotic devices produced significant changes in RF-TIB coupling pattern or variability compared to a no orthotic condition. Therefore, the relief experienced with the INV orthosis was likely due to factors other than alterations in this joint coupling.

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