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EFFECTS OF VARUS ORTHOTICS ON LOWER EXTREMITY KINEMATICS DURING THE PEDAL CYCLE

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ABSTRACT
Purpose. Cycling has been shown to be associated with a high incidence of chronic pathologies. Foot orthoses are frequently used by cyclists in order to reduce the incidence of chronic injuries. The aim of the current investigation was to examine the influence of different varus orthotic inclines on the three-dimensional kinematics of the lower extremities during the pedal cycle. **Methods.** Kinematic information was obtained from ten male cyclists using an eight-camera optoelectronic 3-D motion capture system operating at 250 Hz. Participants cycled with and without orthotic intervention at three different cadences (70, 90 and 110 RPM). The orthotic device was adjustable and four different wedge conditions (0 mm – no orthotic, 1.5 mm, 3.0 mm and 4.5 mm) were examined. Two-way repeated measures ANOVAs were used to compare the kinematic parameters obtained as a function of orthotic inclination and cadence. Participants were also asked to subjectively rate their comfort in cycling using each of the four orthotic devices on a 10-point Likert scale. **Results.** The kinematic analysis indicated that the orthotic device had no significant influence at any of the three cadences. Analysis of subjective preferences showed a clear preference for the 0 mm, no orthotic, condition. **Conclusions.** This study suggests that foot orthoses do not provide any protection from skeletal malalignment issues associated with the aetiology of chronic cycling injuries.

Key words: cycling, biomechanics, foot orthoses

Introduction

Participation in both competitive and recreational cycling has considerably increased as a form of training and also as a leisure activity [1]. However, despite its popularity, cycling has been shown to be associated with a high incidence of chronic pathologies [2, 3]. Chronic musculoskeletal injuries have nonetheless received little attention in cycling research. The few investigations that have been undertaken have unanimously found knee injuries to be the most prevalent complaint, affecting between 24% and 62% of cyclists. Due to the structure of the bicycle and the mechanics of the pedal cycle, the knee joint bears the majority of the load during cycling [4].

Foot orthoses are frequently used by cyclists for a variety of goals [5]. The mechanical reasoning behind the use of foot orthoses is associated with improvements in the biomechanical alignment of the lower extremity and foot, facilitating a more linear cycling motion [6]. This mechanism is considered to be beneficial in preventing chronic injuries in cyclists [7]. The influence of foot orthotic devices has received considerable attention in running biomechanics literature, where orthotic intervention was shown to be an effective treatment of running injuries with a reported success rate of 50–90% [8].

The effects of foot orthoses in cycling has received little attention despite the fact that the foot itself remains one of the primary load-bearing structures in the pedal cycle. Francis [9] proposed that orthotics may be able to compensate for alignment problems in the lower extremities that are linked to the development of injuries. Hannaford et al. [10] utilized an adjustable pedal system to alter foot position in the coronal plane. Although this study did report reductions in self-reported discomfort, the data collection was qualitative only and thus no measurements of the mechanics of the pedal cycle were obtained. Sanderson et al. [11] quantified the effect of a wedge placed between the cycling shoe and pedal on coronal plane kinematics of the knee during steady state cycling. The wedge was able to significantly alter knee coronal plane motion by moving the position of the knee itself away from the bicycle frame. This study utilized two-dimensional video analysis of the knee joint in only one of the three planes of rotation and did not examine the influence of different wedge inclinations on pedal cycle kinematics.

The aim of the current investigation was to examine the influence of different varus wedge inclinations on the three-dimensional (3-D) kinematics of the lower extremities during the pedal cycle. A study of this nature may provide insight into the clinical effectiveness of different foot orthoses and offer better understanding of the mechanism by which orthotic intervention serves to reduce symptoms of chronic cycling injuries. This study tests the hypothesis that orthotic intervention

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will significantly alter the coronal and transverse plane kinematics of the lower extremities, with the larger wedge inclinations having a greater influence.

Material and methods

Ten male cyclists volunteered to take part in this study. Participants were active cyclists training at least three times per week. Basic characteristics of the participants were: age 26.74 ± 2.78 years, height 174.47 ± 4.03 cm and body mass 68.66 ± 4.78 kg. All were free from pathology at the time of data collection and written informed consent was provided in accordance with the Declaration of Helsinki. The procedure was approved by ethics committee of the School of Sport Tourism and Outdoors at the University of Central Lancashire.

Commercially available insoles (High Performance Footbed, Specialized, USA) were utilized in the current investigation. These orthotics feature a varus wedge on their medial aspect and are classified as semi-custom as they allow the extent of the wedge to be altered with three options: 1.5 mm, 3.0 mm and 4.5 mm. Although the right side was selected for analysis, the orthotic devices were placed inside both shoes.

All data were collected using a cycle ergometer (Monark Ergomedic 874E, Monark Exercise, Sweden). Participants were required to cycle with a fixed 2 kg load on the basket at three different cadences of 70, 90 and 110 RPM in each of the four conditions: 0 mm (no orthotic), 1.5 mm wedge, 3.0 mm wedge and 4.5 mm wedge. Saddle height was determined using the LeMond formula [12]. The order in which participants cycled in each of the four orthotic conditions was randomized. Immediately following each trial, participants were asked to rate their subjective comfort in cycling using the orthotic devices using a 10-point Likert scale, with 10 being totally comfortable and 0 being totally uncomfortable.

Kinematic data were obtained using an eight-camera optoelectric motion capture system (Qualisys Medical AB, Sweden) at a capture frequency of 250 Hz. The calibrated anatomical systems technique [13] was used to quantify segmental kinematics. To delineate the anatomical frames of the foot, shank and thigh, retroreflective markers were positioned onto the calcaneus, 1st and 5th metatarsal heads, medial and lateral malleoli, medial and lateral epicondyle of the femur, greater trochanter and iliac crests. To define the pelvic co-ordinate axes, additional markers were placed on the anterior (ASIS) and posterior (PSIS) superior iliac spines. Tracking clusters were also positioned on the shank and thigh segments. A static calibration trial was conducted in which the participant stood in the anatomical position in order for the positions of the anatomical markers to be referenced in relation to the tracking clusters. The hip joint centre was defined using regression modelling based on the position of the ASIS markers [14].

Kinematic curves were time normalized to 100% of the pedal cycle. Movement trials were digitized using Qualisys Track Manager then exported as C3D files. Kinematic parameters were quantified using Visual 3-D software (C-Motion, USA) after marker data were smoothed using a fourth-order zero-lag low-pass Butterworth filter at a cut off frequency of 15 Hz [15]. Three-dimensional kinematics of the hip, knee and ankle joints were calculated using an XYZ cardan sequence referenced to co-ordinate systems about the proximal segment [16]. The designation for rotations were X – sagittal, Y – coronal and Z – transverse plane. Discrete parameters of 1) peak angle during the pedal cycle and 2) relative range of motion (ROM) from top dead centre to peak angle were extracted for statistical analysis.

Descriptive statistics were generated using means and standard deviations for each of the outcome measures. Differences between wedge heights and cadences were examined using two-way repeated measures factorial ANOVA in a 4 × 3 design. In addition, subjective ratings of comfort for each condition were examined using one-way repeated measured ANOVA. Statistical significance was accepted at the $p < 0.05$ level throughout. Appropriate post-hoc analyses were conducted on significant main effects using pairwise comparisons after Bonferroni adjustment to control for type I error. Post-hoc comparisons on significant interactions were conducted using simple main effects analyses. Effect sizes were calculated using partial Eta^2 (pη^2). If the homogeneity assumption was violated then the degrees of freedom were adjusted using the Greenhouse Geisser correction. The Shapiro-Wilk statistic for each condition confirmed that the normal distribution assumption was met in all cases. All statistical procedures were conducted using SPSS 21.0 (SPSS, USA).

Results

A significant main effect was shown for the subjective preferences; $F(3, 27) = 27.68$, $p < 0.05$, p$\eta^2 = 0.74$. Post-hoc analyses showed that each of the four conditions differed significantly from one another, with the highest preferences shown for 0 mm (9.9 ± 0.32), followed by the 1.5 mm wedge (8.3 ± 0.82), 3.0 mm wedge (6.9 ± 0.99) and then 4.5 mm wedge (4.9 ± 0.88).

110 RPM

No significant ($p > 0.05$) differences were observed as a function of the orthotic intervention.

90 RPM

No significant ($p > 0.05$) differences were observed as a function of the orthotic intervention.

70 RPM

No significant ($p > 0.05$) differences were observed as a function of the orthotic intervention.

Table 1. Hip joint kinematics as a function of orthotic wedge inclination and cadence

	110 RPM								90 RPM								70 RPM							
	4.5 mm		3.0 mm		1.5 mm		0.0 mm		4.5 mm		3.0 mm		1.5 mm		0.0 mm		4.5 mm		3.0 mm		1.5 mm		0.0 mm	
	Mean	*SD*	Mean	*SD*	Mean	*SD*	Mean	*SD*	Mean	*SD*	Mean	*SD*	Mean	*SD*	Mean	*SD*	Mean	*SD*	Mean	*SD*	Mean	*SD*	Mean	*SD*
Sagittal plane																								
Peak angle	48.97	18.42	48.78	19.66	44.61	15.23	44.85	15.55	45.35	18.56	47.32	17.12	43.18	15.56	46.53	16.96	43.73	18.19	44.86	16.24	44.28	18.20	39.98	16.98
Relative ROM	44.20	6.35	45.59	3.09	45.51	3.62	44.59	4.03	45.66	6.87	44.72	4.09	46.15	4.25	42.90	5.12	47.40	3.77	44.43	5.11	43.42	4.36	45.30	3.69
Coronal plane																								
Peak angle	–12.65	3.91	–12.46	3.94	–12.32	4.89	–12.91	4.99	–14.29	6.69	–13.78	4.10	–11.47	4.15	–13.01	5.32	–11.71	4.12	–11.34	4.79	–11.59	3.99	–13.67	3.05
Relative ROM	5.26	3.96	4.58	2.88	2.89	1.91	2.59	2.07	3.98	2.82	5.47	3.06	2.34	1.58	2.73	1.45	4.42	3.67	3.62	2.11	2.27	1.22	3.12	1.86
Transverse plane																								
Peak angle	–12.76	12.36	–9.27	9.16	–11.82	10.27	–11.70	10.63	–11.17	9.09	–8.21	8.88	–11.27	9.90	–10.33	9.81	–11.55	7.84	–11.05	10.14	–11.94	10.71	–11.72	9.05
Relative ROM	7.65	3.49	9.48	3.15	7.47	1.20	7.48	1.69	9.92	7.08	10.31	3.80	6.84	0.62	7.21	1.58	9.43	3.33	9.77	1.98	7.99	1.93	10.74	3.74

No significant ($p > 0.05$) differences were observed as a function of the orthotic interventions at a cadence of 110 RPM.

Table 2. Knee joint kinematics as a function of orthotic wedge inclination and cadence

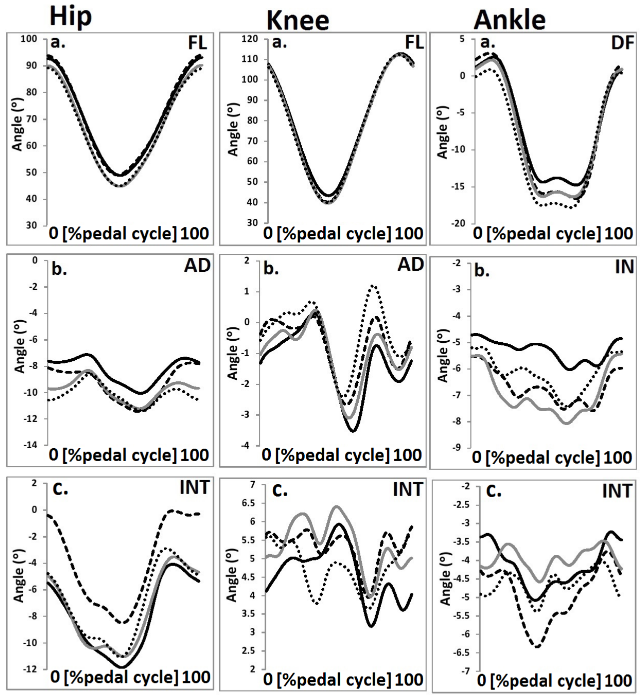
	110 RPM								90 RPM								70 RPM							
	4.5 mm		3.0 mm		1.5 mm		0.0 mm		4.5 mm		3.0 mm		1.5 mm		0.0 mm		4.5 mm		3.0 mm		1.5 mm		0.0 mm	
	Mean	*SD*	Mean	*SD*	Mean	*SD*	Mean	*SD*	Mean	*SD*	Mean	*SD*	Mean	*SD*	Mean	*SD*	Mean	*SD*	Mean	*SD*	Mean	*SD*	Mean	*SD*
Sagittal plane																								
Peak angle	43.01	6.31	40.03	6.49	39.36	6.54	39.53	6.69	40.52	6.95	39.60	6.13	38.16	6.70	42.21	10.35	38.80	7.26	37.85	7.14	40.39	8.03	36.69	6.54
Relative ROM	65.05	7.95	67.52	7.39	67.96	7.71	66.95	8.29	67.99	6.07	67.92	7.20	69.15	7.23	64.82	9.23	70.38	6.64	69.05	7.51	67.21	7.58	71.05	7.56
Coronal plane																								
Peak angle	–4.86	6.36	–4.09	7.53	–4.13	7.68	–3.88	7.16	–5.04	6.35	–4.51	8.03	–4.47	7.54	–2.99	7.56	–5.70	6.45	–4.04	7.60	–4.12	7.42	–4.14	6.88
Relative ROM	3.90	1.79	4.14	1.84	3.47	2.06	3.66	1.95	3.81	2.26	4.22	2.67	3.72	2.25	3.01	1.49	4.55	2.53	4.45	2.53	4.05	1.92	4.52	2.15
Transverse plane																								
Peak angle	–1.68	6.79	–1.60	6.32	–1.70	6.98	–1.89	7.59	–0.06	6.32	–0.54	7.46	–0.53	6.73	–0.71	7.72	–0.41	6.12	0.03	6.41	0.62	6.27	–0.27	6.86
Relative ROM	5.82	3.09	7.32	5.90	6.76	4.29	7.48	5.43	6.09	3.49	6.84	6.62	6.07	4.38	7.03	5.70	6.25	4.11	6.95	5.83	6.60	4.93	8.07	5.22

No significant ($p > 0.05$) differences were observed as a function of the orthotic intervention at a cadence of 90 RPM.

Table 3. Ankle joint kinematics as a function of orthotic wedge inclination and cadence

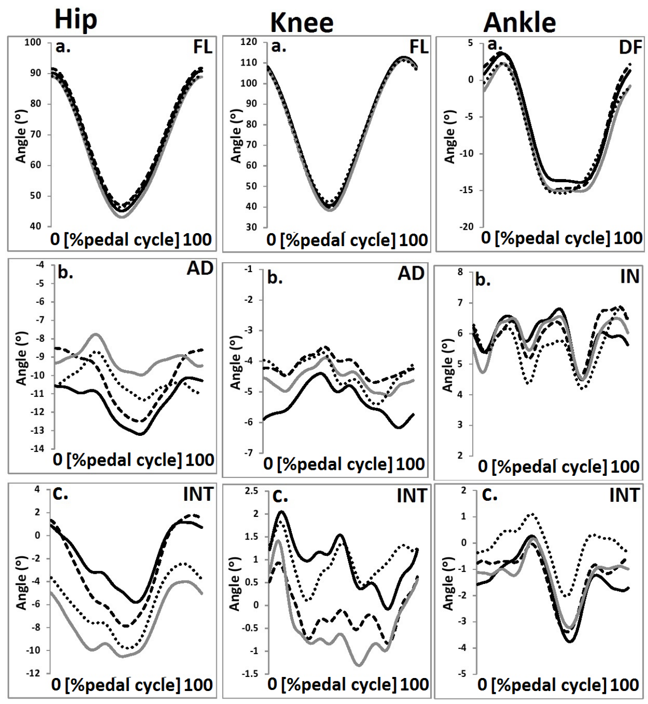
	110 RPM								90 RPM								70 RPM							
	4.5 mm		3.0 mm		1.5 mm		0.0 mm		4.5 mm		3.0 mm		1.5 mm		0.0 mm		4.5 mm		3.0 mm		1.5 mm		0.0 mm	
	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD
Sagittal plane																								
Peak angle	−19.10	5.99	−19.95	7.16	−19.34	7.69	−20.44	10.02	−17.16	6.68	−18.14	8.58	−17.70	8.52	−17.04	10.99	−18.87	8.00	−18.99	9.67	−17.47	9.65	−15.61	10.27
Relative ROM	20.62	5.19	22.69	6.06	20.73	6.16	20.74	8.01	18.34	7.19	20.58	11.33	16.78	8.01	17.04	8.97	22.67	10.33	23.00	11.71	18.87	9.22	18.16	9.73
Coronal plane																								
Peak angle	−6.80	3.42	−6.74	2.57	−7.36	3.38	−6.58	3.93	−3.23	3.37	−1.99	2.22	−2.71	3.68	−2.23	3.70	−3.61	3.52	−2.08	2.18	−3.64	3.13	−3.31	3.67
Relative ROM	1.93	1.14	2.63	1.72	3.00	1.40	2.92	1.43	2.57	1.59	2.09	1.23	1.78	0.79	1.68	0.95	2.03	1.48	2.16	1.30	1.94	0.85	2.10	1.16
Transverse plane																								
Peak angle	−0.57	1.96	−2.29	3.16	−2.45	2.63	−1.95	2.99	−1.09	2.11	−2.17	2.98	−2.72	2.95	−1.38	2.60	−1.20	1.53	−2.16	2.65	−2.50	2.49	−1.50	2.41
Relative ROM	3.09	1.11	4.02	2.49	4.07	1.59	3.68	1.45	2.36	0.48	2.82	1.27	3.62	1.12	2.85	1.00	2.61	1.15	2.67	0.73	3.09	0.75	2.74	0.62

No significant ($p > 0.05$) differences were observed as a function of the orthotic intervention at a cadence of 70 RPM.



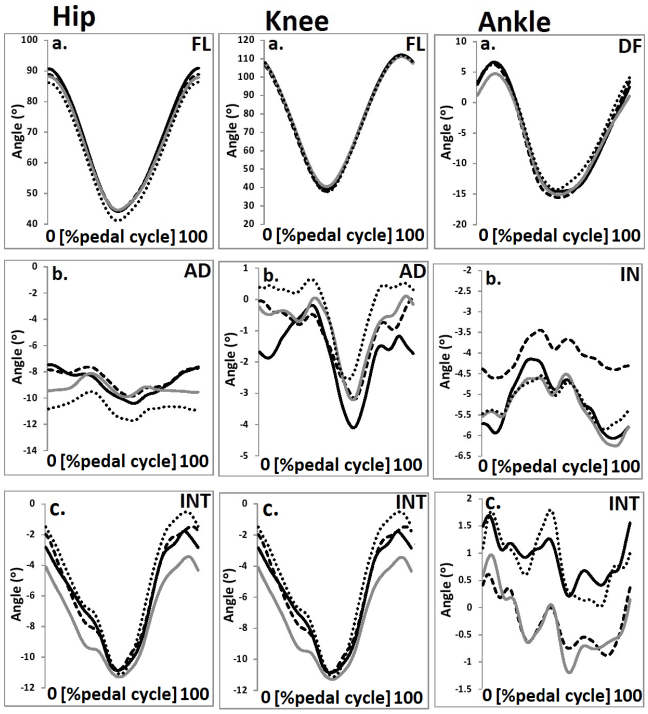
black line – 4.5 mm, dash line – 3.0 mm, grey line – 1.5 mm, dotted line – 0 mm;
FL – flexion, DF – dorsiflexion, AD – adduction, IN – inversion, INT – internal

Figure 1. Hip, knee and ankle joint kinematics at 110 RPM as a function of orthotic condition in the a. – sagittal, b. – coronal and c. – transverse planes



black line – 4.5 mm, dash line – 3.0 mm, grey line – 1.5 mm, dotted line – 0 mm;
FL – flexion, DF – dorsiflexion, AD – adduction, IN – inversion, INT – internal

Figure 2. Hip, knee and ankle joint kinematics at 90 RPM as a function of orthotic condition in the a. – sagittal, b. – coronal and c. – transverse planes



black line – 4.5 mm, dash line – 3.0 mm, grey line – 1.5 mm, dotted line – 0 mm;
FL – flexion, DF – dorsiflexion, AD – adduction, IN – inversion, INT – internal

Figure 3. Hip, knee and ankle joint kinematics at 70 RPM as a function of orthotic condition in the a. – sagittal, b. – coronal and c. – transverse planes

Discussion

The aim of the current investigation was to examine the influence of different varus wedge inclinations on the 3-D kinematics of the lower extremities during the pedal cycle. This investigation represents the first study to examine the influence of varus foot orthotics of different inclincations on the kinematics of the pedal cycle.

The key finding from the current investigation was that orthotic devices, regardless of the wedge incination, did not significantly influence lower extremity kinematics during the pedal cycle. This observation does not support out hypothesis and also opposes the findings of by Sanderson et al. [11], who documented that varus orthotics were able to significantly influence the coronal plane motion of the knee during the pedal cycle and have the desired effect of moving the knee away from the bicycle frame. There are several mechanisms that may explain this discrepancy. Firstly, Sanderson et al. [11] utilized a wedge inclination of 10 mm which is considerably greater than any of the conditions used in the current investigation. An increased wedge inclination is likely to have an enhanced effect on lower extremity kinematics hence the observations of Sanderson et al. [11] are to be expected. Furthermore, the orthotic device may not have been sufficiently rigid to restrain the motion of the ankle joint, meaning that the wedge was not able to successfully attenuate non-sagittal motion during the pedal cycle.

Foot orthoses are frequently advocated for the management of chronic cycling injuries [17], often based on the notion they reduce non-sagittall cycling motions [18]. The observations from the current investigation provide clinically relevant data to cyclists that oppose this notion. Aetiological analyses have confirmed that excessive motions of the hip and knee joints in the coronal and transverse planes are assocaited with aetiology of a number of chronic pathologies in cyclists [5]. Therefore, given that foot orthoses regardless of angulation did not influence lower extremity kinematics, the findings from the current investigation contradict the notion that shoe footbeds serve to reduce the kinematic parameters linked to the aetiology of chronic cycling injuries.

A further important finding from the current investigation is that participants rated the no-orthotic condition as being most preferable for riding comfort followed incremantally by the 1.5, 3.0 and then 4.5mm conditions. This finding, whilst subjective, indicates that riders perceive foot orthoses negatively in terms of their own comfort when pedalling. Furthermore, on the basis that foot orthoses do not appear to provide any clinically beneficial alterations in pedalling mechanics, the current investigation provides evidence indicating that foot orthoses may be unneccesary.

A limitation of the current investigation is that only male cyclists were examined during data collection. This may limit the generalizability of the findings to female cyclists as females are likely to exhibit distinct lower extremity kinematics than males, particularly in the coronal and transverse planes. Therefore, the influence of foot orthoses on the kinematics of the lower extremities during the pedal cycle may be different. It is recommended that the current investigation be repeated using a sample of female cyclists.

Conclusions

The current investigation provides new information on the influence of orthotic foot inserts on the 3-D kinematics of the lower extremities during the pedal cycle. On the basis that no significant alterations in cycling kinematics were observed with the utilization of various foot orthotics, the current investigation may provide insight into the clinical efficacy of orthotic intervention. This study suggests that foot orthoses do not provide any protection from skeletal malalignment issues associated with the aetiology of chronic cycling injuries.

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