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Research article

Effects of Knee Alignments and Toe Clip on Frontal Plane Knee Biomechanics in Cycling

Guangping Shen [1], Songning Zhang [2]✉, Hunter J. Bennett [3], James C. Martin [4], Scott E. Crouter [2] and Eugene C. Fitzhugh [2]

[1] Beijing Pi Fitness Technology Co., Ltd Beijing, China; [2] Department of Kinesiology, Recreation and Sport Studies, The University of Tennessee, Knoxville, TN, USA; [3] Department of Human Movement Sciences, Old Dominion University, Norfolk, VA, USA; [4] Department of Exercise and Sport Science, The University of Utah, Salt Lake City, Utah, USA

Abstract

Effects of knee alignment on the internal knee abduction moment (KAM) in walking have been widely studied. The KAM is closely associated with the development of medial knee osteoarthritis. Despite the importance of knee alignment, no studies have explored its effects on knee frontal plane biomechanics during stationary cycling. The purpose of this study was to examine the effects of knee alignment and use of a toe clip on the knee frontal plane biomechanics during stationary cycling. A total of 32 participants (11 varus, 11 neutral, and 10 valgus alignment) performed five trials in each of six cycling conditions: pedaling at 80 rpm and 0.5 kg (40 Watts), 1.0 kg (78 Watts), and 1.5 kg (117 Watts) with and without a toe clip. A motion analysis system and a customized instrumented pedal were used to collect 3D kinematic and kinetic data. A 3 × 2 × 3 (group × toe clip × workload) mixed design ANOVA was used for statistical analysis ($p < 0.05$). There were two different knee frontal plane loading patterns, internal abduction and adduction moment, which were affected by knee alignment type. The knee adduction angle was 12.2° greater in the varus group compared to the valgus group ($p = 0.001$), yet no difference was found for KAM among groups. Wearing a toe clip increased the knee adduction angle by 0.95º ($p = 0.005$). The findings of this study indicate that stationary cycling may be a safe exercise prescription for people with knee malalignments. In addition, using a toe clip may not have any negative effects on knee joints during stationary cycling.

Key words: Abduction moment, osteoarthritis, knee, workload, varus, valgus.

Introduction

Cycling has advantages in reducing the knee joint loads compared to walking (D'Lima et al., 2008; Kutzner et al., 2012), thus it is frequently prescribed as a rehabilitation exercise by many health professionals (Naal et al., 2007; Salacinski et al., 2012; Walker et al., 2015). Despite the relatively lower joint load during cycling, the prevalence of chronic cycling injuries can be as high as 85% due to poor preparation, equipment and technique as well as overuse (Dettori and Norvell, 2006; Wanich et al., 2007). Among the joints of the lower limb, the knee is thought to be the most affected site with injury prevalence of 42% – 65% (Conti-Wyneken, 1999; Dannenberg et al., 1996; Wilber et al., 1995).

The internal knee abduction moment (KAM) is a surrogate measure for loading to the medial compartment of the knee joint and has been shown to be closely associated with the development of medial knee osteoarthritis (OA) in walking (Mundermann et al., 2004; Sharma et al., 1998; Zhao et al., 2007). Studies have shown that the frontal plane knee malalignment can significantly affect KAM during walking in healthy populations (Barrios et al., 2009; Stief et al., 2011) and knee OA patients (Messier et al., 2014; Turcot et al., 2013). Furthermore, longitudinal studies have shown that varus and valgus alignment were associated with incident and progression of medial and lateral knee OA (Felson et al., 2013; Sharma et al., 2010; Sharma et al., 2001).

Although many studies have investigated the effects of knee alignment during walking, there are a few that have used cycling. Recently, Gardner et al. (2015) compared the KAM in patients with medial knee OA and healthy controls during stationary cycling and found no significant difference between groups. However, knee alignment of the participants was not measured in the study, and it is possible that the knee alignment data may help explain their results on KAM.

Many stationary bikes in fitness or physical therapy facilities have toe clips available and they are used to constrict feet on the pedals during cycling. However, previous studies have suggested allowing some freedom between the foot and pedal may be beneficial for reducing overuse knee injuries (Boyd et al., 1997). It is still unclear whether a toe clip would have any negative effects on knee biomechanics and its effect in individuals with different knee alignment during stationary cycling.

Despite the importance of knee alignment, no studies have explored its effect on knee frontal plane biomechanics during cycling. It is reasonable to assume that the knee alignment may have a similar influence on knee biomechanics during cycling. Furthermore, it remains unclear whether using a toe clip would negatively influence the frontal plane loading in the knee joints. Therefore, the purpose of this study was to examine the effects of knee alignment and use of a toe clip on the knee frontal plane biomechanics during stationary cycling. It was hypothesized that 1) participants with a varus alignment will have a greater KAM compared to participants with a neutral or valgus alignment, and 2) KAM will not differ regardless usage of a toe clip.

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Methods

Participants

Thirty-two participants (11 varus alignment, 11 neutral alignment, and 10 valgus alignment) participated in the study (Table 1). Participants were recruited from a campus student population through flyers and announcements that were made in Kinesiology classes and in Physical Education and Activity Program classes. Using an effect size of 1.091 calculated from the external knee adduction moments in the study by Barrios et al. (Barrios et al., 2009), a total sample size of 10 (or 5 per group) was estimated with a β level of 0.80 and α level of 0.05 (3.1.3, G*Power) (Faul et al., 2007). An anteroposterior full limb radiograph was obtained to measure the knee mechanical axis angle (MAA). Neutral, valgus, and varus groups were determined as 180°±2, >182°, and <178° of knee MAA, respectively (Sharma et al., 2010). The exclusion criteria included a body mass index (BMI) greater than 35 kg/m^2, any injury within the past three months, and inability to ride a stationary bike for 15 minutes. Participants read and signed an informed consent that was approved by the University Institutional Review Board.

Instrumentation

All potential participants attended a full-limb radiographic measurement session. The anteroposterior view of a full-length lower extremity weight-bearing radiograph was captured with the graduated-grid x-ray cassette (Moreland et al., 1987; Sharma et al., 2001). The cassette size was 130.0 cm (height) by 36.0 cm (width). The participant stood barefoot with knees in full extension and the tibial tubercles facing the x-ray beam. The x-ray tube was placed at a distance of 1.83 m from the cassette. The x-ray power settings were 95 kilovolts and 300 mA/s – 500 mA/s, depending on the limb size and tissue characteristics.

A nine-camera motion analysis system (240 Hz, Vicon Motion Analysis Inc., UK) was used to obtain three dimensional (3D) kinematics during the test. Reflective anatomical markers were placed bilaterally on the acromion processes, iliac crests, anterior superior iliac spine, posterior superior iliac spines, greater trochanters, medial and lateral epicondyles, medial and lateral malleoli, 1st and 5th metatarsal heads, tip of the second toe, and midpoint of the front edge of both pedals. A cluster of four tracking markers on a thermoplastic shell was attached to the shanks, thighs, pelvis and trunk. Three discrete markers were also attached to the lateral, superior and inferior heel counters of the standard lab shoes (Noveto, Adidas). Three lateral pedal markers were also used as tracking markers for both pedals (Figure 1) (Fang et al., 2016). A crank tracking marker was placed on the axes of both crank arms, and an additional tracking marker was placed on the front body of the bike (Fang et al., 2016; Gardner et al., 2015).

A mechanically-braked Monark cycle ergometer (818E, Monark, Sweden) was used in this study. The saddle height on the bike was set so that the angle of the knee joint was approximately 30° when the crank was at bottom dead center (Bini et al., 2011). The anterior-posterior position of saddle was set so that the knee was in line with the pedal spindle by a meter stick when the crank was in the forward horizontal position (Burke, 2003). The position of the handlebars was set so that the angle between the participant's trunk and thigh was 90 ° when the crank was in the forward horizontal position (Fang et al., 2016; Gardner et al., 2016).

A customized instrumented bike pedal was used on the cycle ergometer, which allows recordings of three dimensional pedal reaction forces (PRF) and moments (Fang et al., 2016; Gardner et al., 2015; Martin and Brown, 2009). The assembly contained two 3D force sensors (Type 9027C, Kistler, Switzerland) coupled with two industrial charge amplifiers (Type 5073A, Kistler, Switzerland). The charge amplifiers were necessary to convert the charge measured by the force sensors to a voltage value used by the Vicon Nexus software. The sensors were placed in the left pedal and a dummy pedal of the same mass and design was used on the right side to minimize asymmetries during the testing. Pedal analog data along with marker data were exported to Visual3D for further analyses. To align the pedal coordinate system with the lab coordinate system, a customized jig was used to secure the front support base of the bike onto a force platform so that the y-axis of the pedal coordinate system was set parallel to the y-axis of the global coordinate system (Figure 1)(Fang et al., 2016).

Experimental protocol

Participants performed a 2-min warm-up on the cycle ergometer followed by a 2-min rest period. Participants then cycled for 2-min in each of six cycling conditions with a 2-min break between conditions: pedaling at a cadence of 80 rpm and workloads of 0.5 kg (40 Watts), 1.0 kg (78 Watts), and 1.5 kg (117 Watts) with and without a toe clip. All the conditions were randomized by toe clip conditions first, and followed by workload conditions. Simultaneous recordings of kinematic (240 Hz) and kinetic (1200 Hz) data were performed on five consecutive pedal cycles which began during the last 30 seconds of each test condition (Fang et al., 2016; Gardner et al., 2015).

Figure 1. The stationary cycle ergometer setup and marker placements on the pedal and lower limb. The instrumented pedal is on the left side.

Data and statistical analyses

The obtained radiographs were analyzed using InteleViewer software (Intelerad, Montreal, Quebec, Canada). A 2.54 cm diameter sphere was used to calibrate radiographs. The mechanical axis of each limb was then determined using the following standard procedures (Moreland et al., 1987). The mechanical axis of the femur was measured by a line drawn from the center of the femoral head to the center of the tibial intercondylar eminence and the mechanical axis of the tibia was from the center of the intercondylar eminence to the center of the talus (Bennett et al., 2017a; 2017b). The mechanical axis angle of the knee joint was measured by the medial angle between the mechanical axes of femur and tibia. Two investigators (HB and GS) independently performed the same measurements on each radiograph. Inter-rater reliability, as measured by intra-class correlation (ICC), showed that the average ICC was 0.998 with a 95% confidence interval of 0.996 - 0.999 ($F(33,33) = 4982.280$, $p < 0.001$).

Pedal reaction forces (PRF), moments of force, and center of pressure (COP) on the left pedal were computed in Visual 3D (C-Motion Inc.). Pedal reaction forces (PRF) were computed as the respective sums of the vertical (Z), anteroposterior (Y) and mediolateral (X) components of the 3D force sensors. The moments were computed based the PRFs and distance measurements between the two force sensors and of the pedal. The mediolateral center of pressure (COP) was computed based on the equations provided by the manufacturer. The anteroposterior COP displacement was assumed to be fixed along the pedal spindle. The computed PRFs, moments and COP were then transformed into the lab coordinate system for inverse dynamics calculations. The hip joint center was estimated using the Bell method (Bell et al., 1989). A right-hand rule was used to determine the polarity of the joint angles and moments and an X-y-z Cardan rotation sequence was used to compute joint angles with the +Y axis directed anteriorly, the +Z axis superior, and the +X axis orthogonal to the plane of progression and directed laterally to the right. The motion capture data of the last 30 seconds were broken into five movement cycles/trials for further analyses. In cycling conditions, the crank movement cycle of movement trials was defined from the top dead center (0°) to the following top dead center (360°) and angular kinematic and kinetic variables were normalized to the crank cycle of 360°. Raw kinematic and pedal analog signals were filtered using a fourth-order zero-lag Butterworth low pass filter with a cutoff frequency of 6 Hz (Gardner et al., 2015). Customized computer programs (VB_V3D and VB_Table, MS VisualBASIC 6.0) were utilized to identify peak values of selected variables and determine other related variables of the five movement trials for each condition, and organize them for statistical analyses and reports. Pedal reaction forces and joint moments during cycling were not normalized by the participant's body mass or weight as the participant placed most of their body weight on both the seat and handlebars (Fang et al., 2016; Gardner et al., 2015).

A 3 × 2 × 3 (group × toe clip × workload) mixed design analysis of variance (ANOVA) was used to examine the effect of knee alignment, toe clip, and workload on selected biomechanical variables (IBM SPSS Statistics 22, Chicago, IL). When a three-way interaction was found, follow-up two-way ANOVAs were performed. When a two-way ANOVA was significant, post hoc analyses with Bonferroni adjustments were performed to detect specific differences. An alpha level of 0.05 was set a priori. Since the effects of workload on frontal-plane knee biomechanics during stationary cycling have been demonstrated in the previous study (Fang et al., 2016), the emphasis of this study would be placed on the effects of knee alignment and toe clip usage. Therefore, interactions other than group × toe clip interaction were not considered in the discussion.

Results

No significant differences were found for age, height, weight or BMI between the groups (Table 1). The knee MAAs were different among the three alignment groups (all $p < 0.001$, Table 1).

Table 1. Demographic characteristics of participants (mean ± SD).

	Varus (n=11)	Neutral (n=11)	Valgus (n=10)
Age (years)	24.0 ± 2.8	24.0 ± 4.1	22.0 ± 1.6
Height (m)	1.78 ± 0.07	1.76 ± 0.10	1.73 ± 0.07
Weight (kg)	75.1 ± 16.5	73.1 ± 15.3	69.4 ± 7.9
BMI (kg/m^2)	23.6 ± 4.6	23.4 ± 2.9	23.3 ± 2.2
Knee MAA (deg)*	174.3 ± 1.4^{ac}	179.2 ± 1.0^{b}	183.5 ± 1.0

* significant group difference, [a] significantly different between varus and neutral, [b] significant difference between neutral and valgus, [c] significant difference between varus and valgus. BMI: body mass index, MAA: mechanical axis

Pedal reaction force

Workload was significant for peak vertical PRF and medial PRF and both were increased from 0.5 to 1.0 kg, 1.0 to 1.5 kg, and 0.5 to 1.5 kg (all $p < 0.001$, Table 2). Two different knee frontal plane loading patterns, knee abduction and adduction moment, were observed. Ten subjects (out of 11) from the varus group, eight (out of 11) from the neutral group, and five (out of 10) from the valgus group showed an internal knee abduction moment, while the rest of the participants showed an internal knee adduction moment (Figure 2). A Chi-Square test of goodness-of-fit was performed and the distribution of the frontal-plane knee moment patterns was different among alignment groups, χ^2 (2, 32) = 9.28, $p < 0.05$, with 73% exhibiting an internal abductor moment.

Knee joint moment and angle

Workload main effect was significant for peak knee abduction moment, knee extension moment and knee internal rotation moment (all $p < 0.001$). All three variables were increased from 0.5 to 1.0 kg, 1.0 to 1.5 kg, and 0.5 to 1.5 kg (all $p \leq 0.001$, Table 2). A three-way interaction was found for peak knee internal rotation moment ($p = 0.003$, Table 2). Post hoc ANOVAs showed a group × toe clip interaction ($p = 0.007$) only at workload of 0.5 kg. Further analyses showed that the use of toe clip decreased the peak knee internal rotation moment in the varus group ($p = 0.014$) but increased it in the neutral group ($p = 0.039$) at workloads of 0.5 kg only. The main effects for the toe clip and workload were significant for knee extension ROM (p

< 0.04, Table 3). Post hoc results showed that for the toe clip main effect, the extension ROM was 0.42º greater without a toe clip than with a toe clip, and for the workload main effect, knee extension ROM was 0.54º greater at a workload of 1.0 kg compared to 0.5 kg (p = 0.042, Table 3). Group (p = 0.001), toe clip (p = 0.005) and workload (p = 0.035) main effects were significant for peak knee adduction angle (Table 3). Peak knee adduction angle was 12.2° greater in the varus compared to the valgus group (p = 0.001). Toe clip increased peak adduction angle by 0.95º (p = 0.005). Peak knee abduction angle was greater for the valgus compared to the varus group (p < 0.001). Toe clip reduced peak abduction angle by 0.6º (p = 0.008).

A group × toe clip interaction was observed for peak ankle inversion moment (p = 0.021, Table 2). The ankle inversion moment in the toe clip condition was 0.44 Nm greater compared to the no toe clip condition for only varus group (p = 0.001). In addition, there was a toe clip main effect for peak ankle inversion moment (p = 0.044, Table 2) and the moment was greater in the toe clip condition. Workload main effect was significant for ankle inversion ROM (p = 0.047, Table 3), yet no further significant results were revealed in the post hoc analyses. Finally, the use of a toe clip significantly increased the peak foot eversion angle (p < 0.001) and the mean external rotation angle (p < 0.001, Table 3).

Ankle joint moment and angle

Table 2. Peak pedal reaction forces (N) and knee, ankle joint moments (Nm) (mean ± SD).

	Without toe clip			With toe clip		
	0.5 kg	**1.0 kg**	**1.5 kg**	**0.5 kg**	**1.0 kg**	**1.5 kg**
Varus						
Vertical PRF [Yabc]	152.8 ± 31.8	180.3 ± 41.8	236.9 ± 51.1	155.9 ± 42.4	193.4 ± 43.7	236.3 ± 40.6
Medial PRF [Yabc]	21.2 ± 8.9	31.8 ± 9.5	42.8 ± 13.1	16.8 ± 9.6	29.5 ± 16.0	38.7 ± 17.6
Knee Extension Moment [Yabc]	21.4 ± 5.4	33.9 ± 10.0	39.5 ± 9.9	22.8 ± 6.5	31.7 ± 10.3	39.3 ± 10.0
Knee Abduction Moment [%Yabc]	-5.3 ± 1.9	-7.7 ± 2.4	-10.6 ± 4.3	-5.2 ± 2.1	-7.6 ± 4.0	-9.6 ± 3.8
Knee Adduction Moment[$]	6.2 ± 0.0	4.8 ± 0.0	5.3 ± 0.0	4.5 ± 0.0	8.4 ± 0.0	11.8 ± 0.0
Knee Int Rotation Moment[*Y]	4.7 ± 2.2	5.8 ± 3.0	8.6 ± 4.0	3.5 ± 2.5	6.7 ± 4.8	9.9 ± 4.4
Ankle Inversion Moment [#Z]	1.5 ± 1.4	1.6 ± 1.5	1.4 ± 1.3	1.8 ± 1.8	2.0 ± 1.7	2.1 ± 1.6
Neutral						
Vertical PRF	158.2 ± 30.5	199.5 ± 44.7	236.5 ± 44.4	154.1 ± 32.9	192.0 ± 37.2	226.3 ± 49.0
Medial PRF	18.4 ± 7.6	31.4 ± 14.3	41.4 ± 13.4	21.3 ± 9.5	30.1 ± 12.5	37.4 ± 16.2
Knee Extension Moment	23.4 ± 7.4	35.3 ± 10.9	42.5 ± 11.7	26.6 ± 8.0	34.1 ± 11.0	40.3 ± 13.2
Knee Abduction Moment	-4.4 ± 2.3	-7.7 ± 4.3	-8.7 ± 5.2	-5.9 ± 4.2	-6.6 ± 3.9	-8.3 ± 7.1
Knee Adduction Moment	4.2 ± 1.9	4.5 ± 1.5	7.0 ± 3.4	4.3 ± 2.1	5.4 ± 1.9	6.1 ± 1.5
Knee Int Rotation Moment	3.6 ± 1.6	5.6 ± 2.0	7.3 ± 2.1	4.6 ± 2.3	5.4 ± 2.6	6.7 ± 4.4
Ankle Inversion Moment	1.4 ± 0.9	1.8 ± 0.9	2.0 ± 1.4	1.6 ± 1.1	2.0 ± 1.1	2.0 ± 1.4
Valgus						
Vertical PRF	150.7 ± 31.0	183.3 ± 38.1	225.1 ± 43.1	146.9 ± 34.1	191.9 ± 41.5	227.0 ± 47.6
Medial PRF	20.2 ± 6.6	29.3 ± 7.5	38.8 ± 9.9	19.8 ± 4.8	32.6 ± 10.2	38.1 ± 9.9
Knee Extension Moment	22.9 ± 8.0	31.1 ± 8.6	35.6 ± 10.0	22.1 ± 6.3	33.4 ± 8.9	36.3 ± 11.4
Knee Abduction Moment	-3.5 ± 1.3	-5.2 ± 1.2	-6.6 ± 3.0	-4.3 ± 1.1	-5.6 ± 1.2	-6.5 ± 2.0
Knee Adduction Moment	3.7 ± 1.6	4.0 ± 2.0	6.6 ± 5.3	2.7 ± 1.9	3.6 ± 2.1	4.1 ± 2.7
Knee Int Rotation Moment	4.2 ± 2.2	6.0 ± 1.8	8.7 ± 3.8	4.6 ± 1.8	7.9 ± 3.0	9.1 ± 3.4
Ankle Inversion Moment	0.9 ± 0.9	1.0 ± 1.1	1.2 ± 1.8	0.9 ± 1.0	1.1 ± 1.5	0.8 ± 1.1

* significant group x toe-clip x workload interaction; # significant group x toe-clip interaction; X significant group main effect; Y significant workload main effect; Z significant toe-clip main effect. Post hoc comparisons: a significant difference between 0.5 – 1.0 kg, b significant difference between 1.0 – 1.5 kg, c significant difference between 0.5 – 1.5 kg. PRF: Pedal Reaction Force, Mom: moment, Ext.: External, Int.: Internal. % 10 participants from the varus group (10/11), 3 from the neutral group (8/11), and 5 from the valgus group (5/10) showed this pattern. $ 1 subject from the varus group (1/11), 3 from the neutral group (3/11), and 5 from the valgus group (5/10) showed this pattern.

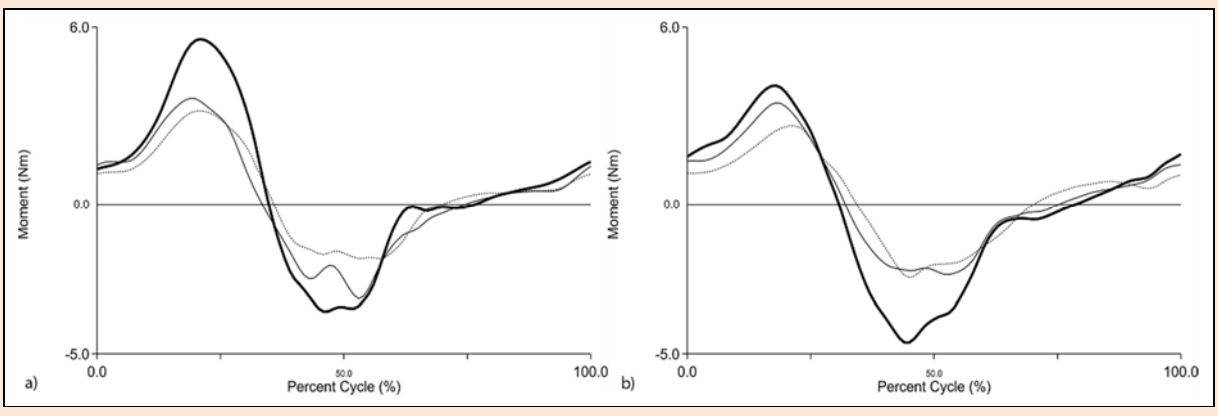

Figure 2. The mean knee adduction moment curves for a) toe clip-free condition and b) toe clip condition. The dotted line is for 0.5 kg workload, the solid line is for 1.0 kg workload, and the bold line is for 1.5 kg workload. Abduction moment is negative.

Table 3. Peak knee, ankle and foot angles and ROM (°) (mean ± SD).

	Without toe clip			With toe clip		
	0.5 kg	1.0 kg	1.5 kg	0.5 kg	1.0 kg	1.5 kg
Varus						
Knee Extension ROM [ZYa]	74.4±5.6	75.1±6.3	74.2±5.4	74.3±5.7	74.1±5.9	74.0±5.5
Knee Adduction Angle [XYZ£]	10.3±4.8	9.7±5.3	10.1±5.2	11.6±4.1	11.0±3.7	10.4±4.3
Knee Abduction Angle [XZ£]	-0.3±4.6	-0.7±5.1	-0.2±4.3	0.3±4.5	0.9±3.3	0.0±4.0
Ankle Eversion Angle	-1.7±7.5	-2.0±6.6	-2.7±6.5	-2.4±6.7	-2.9±7.9	-1.4±6.8
Ankle Inversion ROM[Y]	3.4±1.8	3.6±2.1	4.2±2.5	3.8±2.0	3.7±1.5	3.5±1.6
Foot Eversion Angle[Z]	-3.1±4.2	-2.8±3.7	-2.7±5.5	-5.4±3.7	-5.3±4.4	-4.8±4.6
Foot Mean Ext. Rotation Angle[Z%]	-15.1±4.6	-14.6±5.0	-14.2±4.6	-15.8±3.9	-15.2±4.5	-15.3±3.9
Neutral						
Knee Extension ROM	75.1±6.1	76.2±6.5	76.8±6.0	75.0±6.2	75.6±5.6	75.7±5.8
Knee Adduction Angle	5.2±9.3	5.0±8.8	5.2±8.9	6.0±8.4	5.5±8.7	5.4±9.0
Knee Abduction Angle	-5.0±5.1	-5.2±5.5	-5.6±5.4	-4.7±5.0	-5.0±5.3	-4.8±5.9
Ankle Eversion Angle	-2.8±4.1	-1.8±3.8	-1.1±5.2	-2.4±4.1	-1.4±2.0	-1.4±3.7
Ankle Inversion ROM	3.2±2.1	3.9±2.2	4.0±1.9	2.6±1.6	3.5±2.1	4.2±2.3
Foot Eversion Angle	-4.7±5.3	-2.9±4.9	-2.6±5.2	-7.1±3.9	-5.5±3.5	-4.7±3.3
Foot Mean Ext. Rotation Angle	-12.0±5.0	-11.1±5.5	-11.4±4.6	-13.3±4.1	-12.8±5.1	-12.5±4.7
Valgus						
Knee Extension ROM	73.5±6.4	74.0±5.3	73.4±5.0	73.3±5.2	73.9±6.4	72.8±5.1
Knee Adduction Angle	-2.2±6.4	-2.6±6.7	-2.3±7.0	0.1±7.5	-1.1±6.8	-1.9±6.4
Knee Abduction Angle	-10.4±4.2	-9.7±5.1	-9.9±5.7	-8.9±5.9	-9.6±4.5	-9.6±4.8
Ankle Eversion Angle	-3.6±4.3	-3.8±4.0	-4.0±4.4	-4.1±3.6	-3.5±4.0	-4.0±4.1
Ankle Inversion ROM	4.3±2.4	4.0±2.4	4.4±2.5	3.4±2.3	4.8±3.2	4.3±3.9
Foot Eversion Angle	-3.4±4.3	-3.7±5.0	-4.0±4.0	-6.8±5.1	-6.1±4.9	-6.0±4.2
Foot Mean Ext. Rotation Angle	-10.8±4.0	-10.4±3.9	-11.2±4.4	-13.3±4.0	-12.9±3.0	-12.7±3.7

[X] significant group main effect; [Y] significant workload main effect; [Z] significant toe-clip main effect. Post hoc comparisons: [a] significant difference between 0.5 – 1.0 kg, [b] significant difference between 1.0 – 1.5 kg, [c] significant difference between 0.5 – 1.5 kg. [£] significant difference between varus and valgus. ROM: Range of Motion, Ext.: External, Int.: Internal. [%] Mean angle of first 25% of pedal cycle

Discussion

The purpose of this study was to examine effects of knee alignments and using a toe clip on knee frontal plane biomechanics during stationary cycling. Our first hypothesis that the participants with a varus alignment would have greater KAM compared to those with a neutral or valgus alignment was not supported as no difference in the peak KAM was observed among groups.

Effects of knee alignments on frontal-plane knee biomechanics

It has been suggested that the frontal plane lower limb alignment is highly associated with the magnitude of external knee adduction moment in walking (Barrios et al., 2009; Barrios and Strotman, 2014; Hurwitz et al., 2002; Messier et al., 2014; Stief et al., 2011; Turcot et al., 2013). Considering the significantly different static knee alignments and similar peak medial and vertical PRFs among groups in the current study, it is surprising to see that the KAM did not differ among the three alignment groups. In fact, the gait study that has used the same subject pool in our lab found that the KAM during walking was significantly different among the three alignment groups (Bennett, 2016). Further investigation, however, showed that the varus participants, who displayed an internal knee abduction moment, had the peak knee adduction angle of 10.3 ± 4.6° (n = 10), which did not differ from that of the neutral group (8.6 ± 6.7°, n = 8) or the valgus group (2.7 ±5 .4°, n = 5). As several studies have suggested that the peak knee adduction angle should be more representative of the skeletal geometry of lower extremity during movements (Barrios et al., 2012; Hunt et al., 2008), no difference in the peak knee adduction angle may be partially responsible for the lack of difference of the KAM between the alignment groups in the current study. Another contributing factor may be related to the temporal difference of the peak knee adduction angle and the peak vertical PRF during cycling. The peak knee adduction angle occurred at 23.8° of crank angle (13.2% power phase, Figure 3) and the peak vertical PRF at 116.7° of crank angle (64.9% power phase), whereas the peak KAM occurred at 98.6° of crank angle (54.8% power phase, Figure 4). It is likely that the large temporal separation between the peak knee adduction angle and the peak vertical PRF diminished the effect of knee alignment on the magnitude of KAM. Lastly, not all participants showed an internal knee abduction moment during power phase of cycling, which has been reported previously by Fang et al. (2016). The reduced sample size for KAM may have resulted in lower statistical power to detect the significant difference between the groups.

Interestingly, more participants in the varus group (10 out of 11) showed an internal abduction moment in the power phase than the neutral and valgus groups (8 out of 11 and 5 out of 10, respectively), which suggests that the knee alignment type may play a role in the frontal plane loading patterns during stationary cycling. Besides the knee alignment type, other factors, such as hip kinematics and other lower limb anthropometry (e.g., shank to thigh length ratio), may also influence the frontal-plane knee loading patterns during cycling. Identifying these factors can be helpful for exercise prescription to prevent detrimental knee loading patterns in rehabilitation. Although this is beyond the scope of the present study, further investigations are warranted in these areas of stationary cycling.

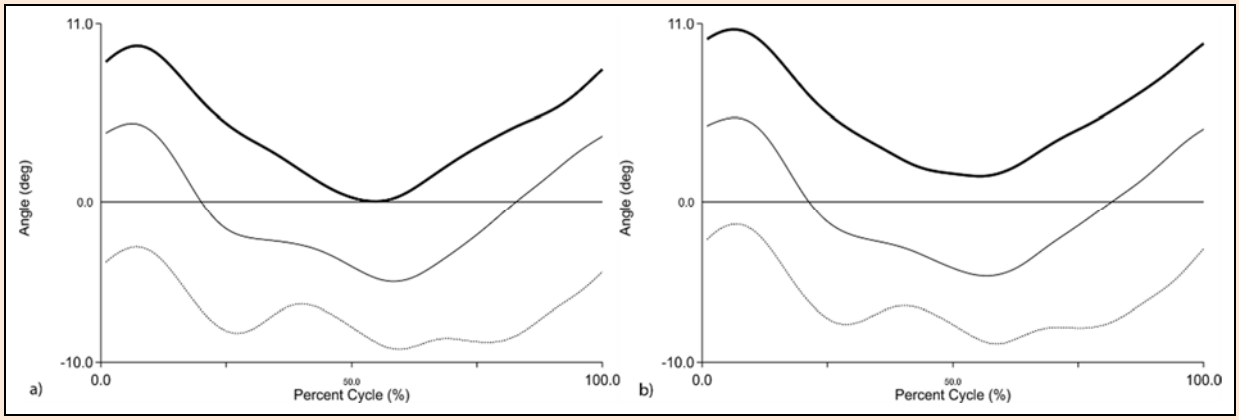

Figure 3. The mean knee adduction angle curves for a) 1.0 kg workload without a toe clip and b) 1.0 kg workload with a toe clip. The bold line is for the varus group; the solid line is for the neutral group; the dotted line is for the valgus group. Adduction angle is positive.

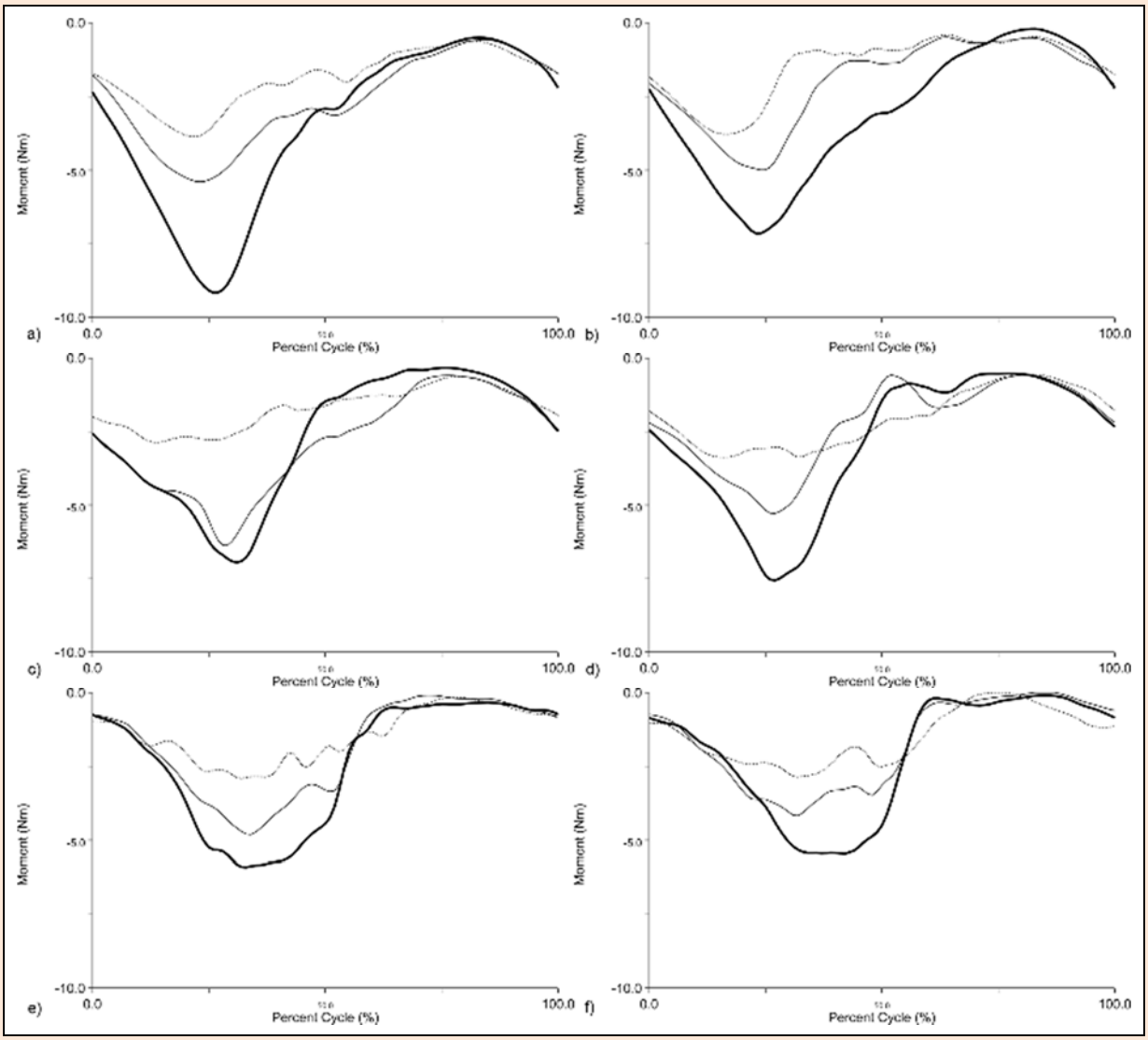

Figure 4. The mean knee abduction moment curves for a) the varus group without a toe clip, b) the varus group with a toe clip, c) the neutral group without a toe clip, d) the neutral group with a toe clip, e) the valgus group without a toe clip and f) the valgus group with a toe clip. The dotted line is for 0.5 kg workload, the solid line is for 1.0 kg workload, and the bold line is for 1.5 kg workload. Abduction moment is negative.

The result of KAMs in cycling is not supported by findings from previous gait studies regarding the effect of knee alignments, which have shown that a static varus alignment is associated with a greater peak knee adduction angle and an increased KAM during walking in both healthy (Barrios et al., 2009; Barrios and Strotman, 2014; Stief et al., 2011) and knee OA populations (Hurwitz et al., 2002; Messier et al., 2014; Turcot et al., 2013). In normal walking, participants have shown a preferred step width ranging from 7.4 to 11.2 cm (Helbostad and Moe-Nilssen, 2003; Hollman et al., 2011; Wert et al., 2010). The KAM was reduced with wider step width in both walking (Fregly, 2008; Zhao et al., 2007) and stair climbing (Paquette et al., 2015; Paquette et al., 2014). The step width in cycling is pre-determined by the distance between the pedals. The distance between the centers of the two pedals in the current study was 31.0 cm, which is similar to the wider step width used in the stair ascent study by Paquette et al. (Paquette et al., 2015), where the average step width of 32.0 cm reduced the first and second peak KAMs by 19.4% and 36.7%, respectively, in healthy participants. Additionally, overall loading to the body reflected in the ground reaction force (GRF) during walking (Barrios et al., 2009; Barrios and Strotman, 2014; Hurwitz et al., 2002; Messier et al., 2014; Stief et al., 2011; Turcot et al., 2013) was much greater than the PRF in cycling. Therefore, it is possible that the greater step width and the smaller PRF resulted in a markedly smaller KAM in cycling compared to the walking, which might have diminished the effect of knee alignments on KAM during stationary cycling. In addition, the peak knee adduction angle has been shown to occur at about the same time (22% of stance) as the peak KAM (23% of stance) during walking (Barrios et al., 2012). It is possible that temporal alignment of the peak adduction angle and GRF is one of the reasons for the varus group having a greater KAM during walking. These results suggest that stationary cycling introduces a less "harmful" frontal-plane movement and loading to the knee joint compared to walking for people with knee malalignment.

The peak KAM increased by 46.1 % and 29.4 % when the workload changed from 0.5 to 1.0 kg and 1.0 to 1.5 kg, respectively. The average peak KAM for all three groups in our study was -7.0 ± 3.0 Nm (0.5%BW×Height) without a toe clip at a workload of 1.0 kg and a cadence of 80 rpm, which is very similar to -7.0±4.3 Nm in the same condition by Fang et al. (2016). These values are also much lower compared to walking, where the peak KAMs are 2.2 - 5.1%BW×Height for knee OA patients and 2.6 - 3.2%BW×Height for healthy controls (Foroughi et al., 2009). Additionally, the peak knee extension moment in the current study did not differ among groups, indicating that the overall knee joint loading during cycling was similar. Therefore, it seems that the stationary cycling is a safe aerobic exercise for people with knee malalignments, including patients with medial knee OA (Sharma et al., 2010). However, further investigations are warranted to examine tibiofemoral contact force in participants with knee malalignment by using the musculoskeletal modeling.

The peak knee abduction angle for the valgus group (-9.7 ± 4.9°) was greater than that of the neutral group (-5.1±5.2°) and the varus group (0.02 ± 4.2°). During stationary cycling, the knee approached peak adduction angle at the beginning (23.8° of crank angle), then abducted through most of the power phase, and reached peak abduction at approximately bottom dead center (171.6° of crank angle, Figure 2). Considering the peak knee abduction angle occurred much later than the peak KAM (75.2° of crank angle), the knee abduction angle was unlikely to have a direct influence on the peak KAM. However, the varus participants still had much smaller peak knee abduction angles which placed them in relatively more adducted knee position during the majority of the power phase compared to the neutral and valgus participants. Although the peak KAM did not differ among groups and the actual magnitudes were much smaller than fully weight bearing exercises, e.g. walking (Fang et al., 2016; Gardner et al., 2015), effects of this more adducted knee angle for people with a varus alignment deserve more attention in future research, especially in the context of long term exercises using stationary cycling.

Effects of using toe clips on knee and ankle biomechanics

The motivation for examining the effects of using a toe clip was to assess if it would have any negative effects on knee biomechanics and how a toe clip would interact with people who have different knee alignments during stationary cycling. Our second hypothesis was supported as no difference in the peak KAM was observed with or without a toe clip during the stationary cycling. It is evident that the small magnitude of change in foot external rotation and eversion angles induced by the toe clip did not make any difference in the KAM during cycling. Furthermore, our result showed that there was no difference in the peak knee extension moment regardless of using a toe clip or not. Based on these results, using a toe clip may not appear to be harmful to the knee mechanics during cycling.

Usage of a toe clip increased the peak foot eversion angle by 2.6° and the mean foot external rotation angle by 1.5°. The toe clip was applied to the foot by tightening the straps between the pedal and the toe clip. A tight toe clip might keep the plantar surface of the foot from lifting away from the pedal, which might have increased foot pronation and minimized the foot inversion, causing the slight increased foot eversion angle. Wearing a toe clip increased the ankle inversion moment by 0.44 Nm only for varus group. It appears that the restricted foot position caused by the toe clip may require a greater inversion moment during the power phase of pedaling, especially for people with a varus alignment.

Another interesting finding was that the use of a toe clip reduced the peak knee abduction angle and increased the peak knee adduction angle during cycling. Perhaps a more everted foot position caused by a toe clip placed the knee joint in a more adducted position, which, however, was not distinctive enough to alter the knee frontal plane loading. An interaction between toe clip and knee alignments was observed for knee internal rotation moment. By using a toe clip, the internal rotation moment was reduced by 1.3 Nm in the varus group and increased by 1.0 Nm in the neutral group compared to the toe clip-free condition at workload of 0.5 kg. It has been reported that the knee rota-

tional moment in gait has been associated with cartilage loss in osteoarthritic knees (Henriksen et al., 2012). Although the participants and the movement in the present study were different than the reported one (Henriksen et al., 2012), our finding of the reduced internal rotation moment suggests that wearing a toe clip may be beneficial for the people with a varus knee alignment during stationary cycling at a lower workload. Additionally, the absolute change caused by the toe clip was quite small. Therefore, caution should be used when interpreting the results of the current study.

Limitations

One limitation of this study is that the participants might have differed in their previous experiences in cycling, which may have introduced greater variability in pedaling techniques and results. However, considering that diverse techniques are usually performed by people who are engaged in stationary cycling as a recreational exercise or rehabilitation, the variability observed in the current study may be more generalizable to the actual cycling characteristics of the population. The mechanical axis was determined from long standing X-ray without consideration of tibial torsion, which may introduce potential bias in determining lower limb alignments.

Conclusions

The findings of this study indicate that a varus or valgus knee alignment did not result in abnormal frontal-plane knee loading, suggesting stationary cycling may be a safe exercise prescription for people with knee malalignments. Using a toe clip did not lead to a greater peak KAM, suggesting it may not have any negative effects on knee joints during stationary cycling. This is the first study that examined the effects of knee alignments and using a toe clip on knee frontal-plane biomechanics during stationary cycling. Future studies should examine the actual tibiofemoral contact force in participants with different knee alignments by using the musculoskeletal modeling.

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The authors thank all participants for their contributions. This study presents no conflicts of interest. The authors declare that the results of the study are presented clearly, honestly, and without fabrication, falsification, or inappropriate data manipulation. The experiments comply with the current laws of US.

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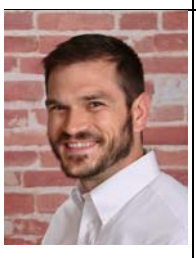
Key points

- Varus or valgus alignment did not cause increased frontal-plane knee joint loading, suggesting stationary cycling is a safe exercise.
- This study supports that using a toe clip did not lead to abnormal frontal-plane knee loading during stationary cycling.
- Two different knee frontal plane loading patterns, knee abduction and adduction moment, were observed during stationary cycling, which are likely affected by the type of knee alignment.

AUTHOR BIOGRAPHY

Guangping SHEN
Employment
CEO, Beijing Pi Fitness Technology Co., Ltd Beijing, China
Degree
MS
Research interest
Knee Alignment, Cycling Biomechanics, Knee Osteoarthritis
E-mail: guangpingshen@gmail.com

Songning ZHANG
Employment
Professor, Department of Kinesiology, Recreation, & Sport Studies, The University of Tennessee, Knoxville
Degree
PhD
Research interest
Gait and cycling biomechanics, knee osteoarthritis, total knee replacement, synthetic turf biomechanics, musculoskeletal modeling
E-mail: szhang@utk.edu



Hunter J. BENNETT
Employment
Assistant Professor and Neuromechanics Lab Director in the Department of Human Movement Sciences, Old Dominion University, Norfolk, VA.
Degree
PhD
Research interest
Biomechanics in sport and clinical gaits, neuromuscular control strategies, and musculoskeletal simulations of human movement
E-mail: hjbennet@odu.edu

James C. MARTIN
Employment
Associate Professor, Department of Nutrition & Integrative Physiology, University of Utah
Degree
PhD
Research interest
Neuromuscular function, biomechanics, musculoskeletal modeling
E-mail: jim.martin@utah.edu

Scott E. CROUTER
Employment
Associate Professor, Department of Kinesiology, Recreation, and Sport Studies, The University of Tennessee Knoxville
Degree
PhD
Research interest
Physical activity measurement with wearable sensors; machine learning algorithms
E-mail: scrouter@utk.edu

Eugene C. FITZHUGH
Employment
Associate Professor, Department of Kinesiology, Recreation, and Sports Studies, University of Tennessee, Knoxville
Degree
PhD
Research interest
Physical activity epidemiology, active living
E-mail: fitzhugh@utk.edu

✉ **Songning Zhang, PhD, FACSM**
Biomechanics/Sports Medicine Lab, Department of Kinesiology, Recreation, & Sport Studies, The University of Tennessee
1914 Andy Holt Avenue, Knoxville, TN 37996-2700, USA