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Creators	Sinclair, Jonathan Kenneth, Butters, Bobbie, Brooks, Darrell and Stainton, Philip

Effects of a prophylactic knee bracing on patellofemoral loading during cycling.

Jonathan Sinclair[1], *Bobbie Butters*[1], *Darrell Brooks*[2] & *Philip Stainton*[1]

1. Centre for Applied Sport and Exercise Sciences, Faculty of Health and Wellbeing, University of Central Lancashire, Lancashire, UK.
2. School of Pharmacy and Biomedical Sciences, Faculty of Clinical and Biomedical Sciences, University of Central Lancashire, Lancashire, UK.

Correspondence Address:

Dr. Jonathan Sinclair

Centre for Applied Sport Exercise and Nutritional Sciences

School of Sport and Wellbeing

Faculty of Health and Wellbeing

University of Central Lancashire

Preston

Lancashire

PR1 2HE.

e-mail: jksinclair@uclan.ac.uk

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Abstract

PURPOSE: The aim of the current investigation was to utilize a musculoskeletal simulation approach to examine the effects of prophylactic knee bracing on patellofemoral joint loading during the pedal cycle.

METHODS: Twenty-four (12 male and 12 female) healthy recreational cyclists rode a stationary cycle ergometer at fixed cadences of 70, 80 and 90 RPM in two different conditions (brace and no-brace). Patellofemoral loading was explored using a musculoskeletal simulation approach and participants were also asked to subjectively rate their perceived stability and comfort whilst wearing the brace.

RESULTS: The results showed that the integral of the patellofemoral joint stress was significantly lower in the brace condition (male: 70RPM=8.89, 80RPM=9.76, & 90RPM=12.30 KPa/kg·s and female: 70RPM=11.59, 80RPM=13.07 & 90RPM=14.14 KPa/kg·s) compared to no-brace (male: 70RPM=10.23, 80RPM=10.96 & 90RPM=13.20 and female: 70RPM=12.43, 80RPM=14.04 & 90RPM=15.45 KPa/kg·s). In addition, it was also revealed that participants rated that the knee brace significantly improved perceived knee joint stability.

CONCLUSIONS: The findings from the current investigation therefore indicate that prophylactic knee bracing may have the potential to attenuate the risk from the biomechanical parameters linked to the aetiology of patellofemoral pain in cyclists. Future, longitudinal analyses are required to confirm the efficacy of prophylactic knee braces for the attenuation of patellofemoral pain symptoms in cyclists.

Introduction

Road cycling has been an Olympic discipline for over 100 years and is regarded as one of the world's most popular sporting events (1). Cycling is associated with a plethora of physiological and psychological benefits and is practiced at both competitive and recreational levels by millions of participants worldwide (2). However, despite being considered a non-weight bearing activity (3), cycling is associated with a high rate of injuries (4).

Patellofemoral pain is the most frequently experienced musculoskeletal condition, affecting 36% of all cyclists and accounting for more than 57 % of all time-loss pathologies (4, 5). Patellofemoral pain is so prevalent in cycling that it has been termed ‘cyclist’s knee’ (6) and the long term forecast for patients is poor, as many later present with radiographic patellofemoral joint osteoarthritis (7). Elevated patellofemoral joint stress is the biomechanical mechanism linked most strongly to the aetiological of patellofemoral pain (8), and although, musculoskeletal modeling approaches exist to estimate patellofemoral joint loading (9, 10), they require inverse dynamics as input parameters into the musculoskeletal algorithm. Joint torques are not representative of localized joint loading, as Herzog et al., (11) showed that muscles are the primary contributors to lower extremity joint kinetics. Recent advances in musculoskeletal simulation software and associated models including the patellofemoral joint (12) have been developed, which allow skeletal muscle force distributions to be simulated during movement and utilized as input parameters for the quantification of lower extremity joint loading. To date, there has been only limited utilization of musculoskeletal simulation for cycling specific analyses.

Given the high incidence of patellofemoral pain in athletic and active populations, a range of conservative prophylactic and treatment modalities have been explored in biomechanical and clinical literature. Prophylactic braces are designed to prevent knee pathologies by reducing the magnitude of the biomechanical mechanisms linked to the aetiology of injury and by enhancing joint proprioception (13). Prophylactic knee braces represent an inexpensive conservative modality, designed to be minimally restrictive during sports tasks (14, 15). Prophylactic knee braces are utilized extensively; yet only one study currently exists exploring the biomechanical effects of knee bracing during cycling. Theobald et al., (16) explored the effects of knee bracing and patella taping on three-dimensional knee joint kinematics during stationary cycling at different workloads. Their findings showed that the brace significantly reduced the coronal plane knee range of motion and also the peak transverse plane angle compared to taping, although their participants revealed that the brace was too uncomfortable to be clinically viable. However, to date, there has yet to be any investigation, which has examined the effects of prophylactic knee bracing on patellofemoral joint loading linked to the aetiology of patellofemoral pain during cycling.

Therefore, the aim of the current investigation was to utilize a musculoskeletal simulation approach to examine the effects of prophylactic knee bracing on patellofemoral joint loading during the pedal cycle. A study of this nature may provide important clinical information regarding the efficacy of knee bracing for the prevention of patellofemoral pain in cyclists. The current investigation tests the hypothesis that prophylactic knee bracing will serve to reduce patellofemoral stress linked to the aetiology of injury.

Methods

Participants

Twenty-four recreational cyclists (12 male and 12 female), volunteered to take part in this study. All had at least 2 years of road cycling experience and were from lower extremity musculoskeletal pathology at the time of data collection. The mean characteristics of the participants were; (males) age 28.14 ± 6.31 years, height 1.77 ± 0.07 m and body mass 79.04 ± 9.25 kg and (females) age 26.71 ± 5.65 years, height 1.64 ± 0.06 m and body mass 62.56 ± 7.33 kg. To be eligible for participation, cyclists were required to have at least 2 years of road cycling experience. In addition, they were required to be free from musculoskeletal pathology at the time of data collection, with no previous knee joint surgical intervention. The procedure utilized for this investigation was approved by the University of Central Lancashire, Science, Technology, Engineering and Mathematics, ethical committee (Ref: 644) and all participants provided written informed consent

Knee brace

A single nylon/silicone knee brace was utilized in this investigation, (Kuangmi 1 PC compression knee sleeve), which was worn on the dominant (right) limb in all participants. The brace examined, as part of this study is lightweight knee joint compression sleeve designed to provide support and enhance joint proprioception.

Procedure

Participants rode a stationary ergometer SRM 'Indoor Trainer' (SRM, Schoberer, Germany) for 6 minutes at fixed cadences of 70, 80 and 90 RPM in both brace and no-brace conditions. The experimental conditions were completed in a counterbalanced order and a standardized rest period of 5 minutes was allowed between trials. The bicycle set-up was conducted in accordance with previous recommendations (17), and maintained between each condition. The cycling shoes (Northwave Sonic 2 Plus Road), pedals (Look Keo Classic 2, Look, Cedex, France), cleats (Look Keo Grip, 4.5˚ float, Look, Cedex, France), chain ring (SRM power, SRM, Schoberer, Germany) and crank (SRM power, SRM, Schoberer, Germany) were also maintained across all trials, and positioned in accordance with previous recommendations (18). Participants were given continuous visual feedback of their cadence, which was visible via the SRM head unit (Powercontrol V, SRM, Schoberer, Germany).

Kinematic information from the lower extremity joints was obtained using an eight camera motion capture system (Qualisys Medical AB, Goteburg, Sweden) using a capture frequency of 250 Hz. To define the anatomical frames of the thorax, pelvis, thighs, shanks and feet retroreflective markers were placed at the C7, T12 and xiphoid process landmarks and also positioned bilaterally onto the acromion process, iliac crest, anterior superior iliac spine (ASIS), posterior super iliac spine (PSIS), medial and lateral malleoli, medial and lateral femoral epicondyles, greater trochanter, calcaneus, first metatarsal and fifth metatarsal. Carbon-fibre tracking clusters comprising of four non-linear retroreflective markers were positioned onto the thigh and shank segments. In addition to these the foot segments were tracked via the calcaneus, first metatarsal and fifth metatarsal, the pelvic segment was tracked using the PSIS and ASIS markers and the thorax segment was tracked using the T12, C7 and xiphoid markers. Static calibration trials were obtained with the participant in the anatomical position in order for the positions of the anatomical markers to be referenced in relation to the tracking clusters/markers. A static trial was conducted with the participant in the anatomical position in order for the anatomical positions to be referenced in relation to the tracking markers, following which those not required for dynamic data were removed.

In addition to the biomechanical movement information, the effects of the experimental brace on knee joint proprioception were also examined using a cycling specific joint position sense test. This was conducted, in accordance with the procedure of Drouin et al., (29), whereby participants were assessed on their ability to reproduce a target knee flexion angle whilst sat on the cycle ergometer. To accomplish this, participants were asked to slowly turn the pedal to 90 ° from the point of top dead centre, which was verified using a handheld goniometer by the same researcher throughout data collection. Participants then held this position for 15 seconds during which time the 'criterion' knee flexion position was captured using the motion analysis system. Following this, participants were asked to pedal at a fixed cadence of 60 RPM for 60 seconds, after which they reproduced the target position as accurately as possible but without guidance via the goniometer. Again, this position was held for a period of 15 seconds and the knee flexion angle during the 'replication' trial was also collected using the motion analysis system. This above process was conducted on three occasions in both the brace and no-brace conditions in a counterbalanced order. The absolute difference in degrees calculated between the criterion and replication trials was averaged over the three trials to provide angular error values in both brace and no-brace conditions, which were extracted for statistical analysis.

Following completion of the biomechanical data collection, in accordance with Sinclair et al., (20), participants were asked to subjectively rate the knee brace in relation to performing the cycling movements without the brace in terms of stability and comfort. This was accomplished using 3 point scales that ranged from 1 = more comfortable, 2 = no-change and 3 = less comfortable and 1 = more stable, 2 = no-change and 3 = less stable.

Processing

Marker trajectories were identified using Qualisys Track Manager, then exported as C3D files to Visual 3D (C-Motion, Germantown, MD, USA). Marker data were smoothed using a cut-off frequency 12 Hz using a low-pass Butterworth 4th order zero-lag filter (20).

All biomechanical data were normalized to 100% of the pedal cycle, which was delineated using concurrent instances in which the right pedal was positioned at top dead centre, in accordance with Sinclair et al., (21). Within Visual 3D, five pedal cycles were obtained during minutes 2-3 of the experimental protocol. Three-dimensional kinematics of the knee were calculated using an XYZ cardan sequence of rotations (where X = sagittal plane; Y = coronal plane and Z = transverse plane). The maximum knee range of motion (representative of the angular difference between maximum and minimum angles during the pedal cycle) in each plane of rotation was extracted for statistical analysis.

Data from the five pedal cycles in each condition were then exported from Visual 3D into OpenSim 3.3 software (Simtk.org). A validated musculoskeletal model with 12 segments, 19 degrees of freedom and 92 musculotendon actuators (12) was used to quantify patellofemoral joint forces. The model was firstly scaled for each participant to account for the anthropometrics of each rider. We firstly performed a residual reduction algorithm (RRA) within OpenSim; in order to reduce the residual forces and moments (22). As muscle forces are the main determinant of joint compressive forces (11), muscle kinetics were quantified using a static optimization process in accordance with Steele et al., (23). Following this patellofemoral, joint forces were calculated using the joint reaction analyses function using the muscle forces generated from the static optimization process as inputs. Finally, patellofemoral joint stress was quantified by dividing the patellofemoral force by the patellofemoral contact area. Patellofemoral contact area were obtained by fitting a 2nd order polynomial curve to the sex specific data of Besier et al., (24), who estimated patellofemoral contact areas as a function of the knee flexion angle using MRI.

All patellofemoral and muscle forces were normalized by dividing the net values by body mass (N/kg). From the above processing, peak patellofemoral force, and peak patellofemoral stress (KPa/kg) were extracted for statistical analysis. Furthermore, the peak forces during the pedal cycle of the muscles crossing the knee joint (rectus femoris, vastus lateralis, vastus medialis, vastus intermedius, biceps femoris long head, biceps femoris short head, semitendinosus, semimembranosus, medial gastrocnemius, lateral gastrocnemius, sartorius and gracilis) were also extracted. In addition, the integral of the patellofemoral joint force (N/kg·s), patellofemoral joint stress (KPa/kg·s) and muscles forces (N/kg·s) were calculated during the pedal cycle using a trapezoidal function. The patellofemoral force instantaneous load rate (N/kg/s) was also extracted by obtaining the peak increase in force between adjacent data points. Finally, the patellofemoral contact area at the instance of peak patellofemoral joint stress and mean contact area during the pedal cycle were also obtained for statistical analysis.

Statistical analyses

Descriptive statistics of means and standard deviations were obtained for each outcome measure. Shapiro-Wilk tests were used to screen the data for normality. Differences in knee proprioception with and without the presence of the brace were examined using a 2 (BRACE) x 2 (GENDER) mixed ANOVA. Differences in biomechanical parameters were examined using 2 (BRACE) x 3 (WORKLOAD) x 2 (GENDER) mixed ANOVA's. In the event of a significant main effect, pairwise comparisons were performed and any significant interactions were explored using simple main effects. In addition, the subjective ratings in relation to the stability and comfort of the knee sleeve were examined using Chi-Squared (X^2) tests. Statistical significance was accepted at the $P \leq 0.05$ level. Effect sizes for all significant findings were calculated using partial Eta^2 ($p\eta^2$). All statistical actions were conducted using SPSS v24.0 (SPSS Inc, Chicago, USA).

Results

Tables 1-6 present the mean ± SD kinetics and kinematics as a function of different brace workload conditions.

Patellofemoral joint kinetics and contact area

For peak patellofemoral force, a significant main effect of WORKLOAD was observed ($P<0.05$, $p\eta^2 = 0.18$). Post-hoc pairwise comparisons showed that peak force was statistically larger in the 90 RPM condition compared to 70 RPM ($P=0.02$) *(Table 1 & 2)*. In addition, for peak patellofemoral stress, a significant main effect of WORKLOAD was shown ($P<0.05$, $p\eta^2 = 0.17$). Post-hoc pairwise comparisons showed that peak force was statistically larger in the 90 RPM condition compared to 70 RPM ($P=0.03$) *(Table 1 & 2).*

For the integral of the patellofemoral joint force, significant main effects of both WORKLOAD ($P<0.05$, $p\eta^2 = 0.14$) and BRACE ($P<0.05$, $p\eta^2 = 0.28$) were noted. Post-hoc pairwise comparisons for WORKLOAD showed that the patellofemoral force integral was statistically larger in the 90 (P=0.04) and 80 RPM (P=0.03) conditions compared to 70 RPM. For BRACE it was shown that the integral of the patellofemoral joint force was statistically larger in the no-brace condition (P=0.008) *(Table 1 & 2)*. In addition, for the integral of the patellofemoral joint stress, a significant main effect of for BRACE (P<0.05, $p\eta^2 = 0.27$) was noted, with the patellofemoral integral stress being statistically larger in the no-brace condition (P=0.009) *(Table 1 & 2)*.

No further statistical differences were observed *(Table 1 & 2)*.

@@@TABLE 1 NEAR HERE@@@

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Muscle kinetics

For the peak rectus femoris force a significant main effect of WORKLOAD (P<0.05, $p\eta^2 = 0.31$) was found. Post-hoc pairwise comparisons showed that peak force was statistically larger in the 90 RPM compared to the 70 (P=0.002) and 80 RPM (P=0.03) conditions and that 80 RPM was larger than 70 RPM (P=0.0004) *(Table 3 & 4)*. For the integral of the rectus femoris force a significant BRACE main effect was found (P<0.05, $p\eta^2 = 0.23$), with the integral force being statistically larger in the no-brace condition (P=0.02) *(Table 3 & 4)*.

For the peak vastus lateralis force, significant main effects of WORKLOAD (P<0.05, $p\eta^2 = 0.18$) and BRACE (P<0.05, $p\eta^2 = 0.21$) were found. Post-hoc pairwise comparisons for

WORKLOAD showed that peak force was statistically larger in the 80 ($P=0.04$) and 90 RPM ($P=0.02$) conditions than 70 RPM. For BRACE the peak force was statistically larger in the no-brace condition ($P=0.02$) *(Table 3 & 4).*

For the peak vastus medialis force, significant main effects of WORKLOAD ($P<0.05$, $p\eta^2 = 0.17$) and BRACE ($P<0.05$, $p\eta^2 = 0.24$) were found. Post-hoc pairwise comparisons for WORKLOAD showed that peak force was statistically larger in the 90 RPM ($P=0.03$) condition than 70 RPM. For BRACE the peak force was statistically larger in the no-brace condition ($P=0.02$) *(Table 3 & 4).* For the integral of the vastus medialis force a significant BRACE main effect was found ($P<0.05$, $p\eta^2 = 0.17$), with the integral force being statistically larger in the no-brace condition ($P=0.04$) *(Table 3 & 4).*

For the peak vastus intermedius force, significant main effects of WORKLOAD ($P<0.05$, $p\eta^2 = 0.17$) and BRACE ($P<0.05$, $p\eta^2 = 0.27$) were found. Post-hoc pairwise comparisons for WORKLOAD showed that peak force was statistically larger in the 90 RPM ($P=0.03$) condition than 70 RPM. For BRACE the peak force was statistically larger in the no-brace condition ($P=0.009$) *(Table 3 & 4).* For the integral of the vastus intermedius force a significant BRACE main effect was found ($P<0.05$, $p\eta^2 = 0.17$), with the integral force being statistically larger in the no-brace condition ($P=0.04$) *(Table 3 & 4).*

For the peak biceps femoris long head force, significant main effects of WORKLOAD ($P<0.05$, $p\eta^2 = 0.29$) and BRACE ($P<0.05$, $p\eta^2 = 0.34$) were found. Post-hoc pairwise comparisons for WORKLOAD showed that peak force was statistically larger in the 80 ($P=0.001$) and 90 RPM

(P=0.004) conditions than 70 RPM (P=0.03). For BRACE the peak force was statistically larger in the no-brace condition (P=0.003) *(Table 3 & 4).* For the integral of the biceps femoris long head force a significant BRACE main effect was found ($P<0.05$, $p\eta^2 = 0.32$), with the integral force being statistically larger in the no-brace condition (P=0.004) *(Table 3 & 4).*

For the peak biceps femoris short head force, a significant main effect of WORKLOAD ($P<0.05$, $p\eta^2 = 0.43$) was found. Post-hoc pairwise comparisons showed that peak force was statistically larger in the 90 RPM compared to the 70 (P=0.00009) and 80 RPM (P=0.003) conditions and that 80 RPM was larger than 70 RPM (P=0.0005) *(Table 3 & 4).*

For the peak semimembranosus force, a significant main effect of WORKLOAD ($P<0.05$, $p\eta^2 = 0.18$) was found. Post-hoc pairwise comparisons showed that peak force was statistically larger in the 90 (P=0.03) and 80 RPM (P=0.02) conditions compared to 70 RPM *(Table 3 & 4)*.

For the peak sartorius force, a significant main effect of WORKLOAD ($P<0.05$, $p\eta^2 = 0.23$) was found. Post-hoc pairwise comparisons showed that peak force was statistically larger in the 90 (P=0.002) and 80 RPM (P=0.008) conditions compared to 70 RPM *(Table 3 & 4)*.

No further statistical differences were observed *(Table 3 & 4)*.

@@@TABLE 3 NEAR HERE@@@

@@@TABLE 4 NEAR HERE@@@

Three-dimensional kinematics

In the sagittal plane, a significant main effect of WORKLOAD ($P<0.05$, $p\eta^2 = 0.20$) was found. Post-hoc pairwise comparisons showed that the sagittal plane maximum knee range of motion (ROM) was statistically larger in the 90 RPM compared to the 70 ($P=0.02$) and 80 RPM ($P=0.006$) conditions *(Table 5 & 6)*.

In the coronal plane, significant main effects of WORKLOAD ($P<0.05$, $p\eta^2 = 0.22$) and BRACE ($P<0.05$, $p\eta^2 = 0.24$) were found. Post-hoc pairwise comparisons showed that the coronal plane maximum knee ROM was statistically larger in the 90 RPM compared to the 70 ($P=0.02$) and 80 RPM ($P=0.02$) conditions (Table 5 & 6). For BRACE maximum coronal knee ROM was statistically larger in the no-brace condition ($P=0.02$) *(Table 5 & 6)*.

No further statistical differences were observed *(Table 5 & 6)*.

@@@TABLE 5 NEAR HERE@@@

@@@TABLE 6 NEAR HERE@@@

Knee proprioception

No significant differences (P>0.05) in knee proprioception were shown. In the no-brace condition, a mean error of 4.70 ± 2.59 ° was found for males and 6.90 ± 4.05 ° shown for females. In the brace condition, a mean error of 3.74 ± 2.58 ° was found for males had and 6.34 ± 3.60 ° shown for females.

Subjective preferences

For comfort the Chi-Squared test was not significant (X^2 = 1.25, P=0.27), with 9 participants rating the brace as more comfortable, 11 as no-change and 4 as less comfortable. For stability however the Chi-Squared test was significant (X^2 = 5.00, P=0.03), with 14 participants rating the brace as more stable, 10 as no-change and 0 as less stable.

Discussion

Patellofemoral pain the most frequent musculoskeletal condition in cyclists (1, 5), with a poor long-term prognosis (7). In support of the hypothesis, the current investigation importantly revealed that in both males and females, the integral of the patellofemoral contact stress was significantly reduced when wearing the brace. This finding may be important regarding the initiation and progression of patellofemoral pain in cyclists, as patellofemoral pain symptoms are mediated through excessive patellofemoral joint stress (8). Therefore, the current investigation indicates that prophylactic knee bracing may have the potential to attenuate the biomechanical parameters linked to the aetiology of patellofemoral pain in cyclists. Nonetheless, it is important to acknowledge that this represents an acute intervention only and longitudinal analyses are required before the above notion can be substantiated.

This investigation also showed that there were no statistical differences in patellofemoral contact area. As stress is a reflection of the joint reaction force divided by the contact area, the reductions in patellofemoral stress were mediated by the corresponding decrease in the integral of the patellofemoral joint reaction force. As the quadriceps is the only muscle to cross the patellofemoral joint, forces produced by this muscle group play a significant role in the generation of compressive reaction forces at this joint (9). Therefore, it is proposed that the attenuation of the patellofemoral joint reaction force in the brace condition was observed primarily due to the significant reductions in the integral of each of the four-quadriceps muscle forces during the pedal cycle. Indeed this notion is supported by those of Besier et al., (25) indicating that patients with patellofemoral pain exhibit increased quadriceps muscle forces in relation to pain free controls.

The significant reduction in peak biceps femoris long head force in the brace condition is an interesting observation. This finding agrees with the assertions of Elias et al., (26), indicating that the hamstring muscle group contributes to patellofemoral joint loading. Such increases in hamstring force production may mediate posterior translation of the tibia (27). This serves to attenuate the effective moment arm of the quadriceps (28), resulting in a compensatory increase in quadriceps force. Enhanced hamstring muscle forces may also provide resistance to knee extension given the high levels of knee flexion typically associated with cycling (27). The hamstring group and biceps femoris muscle in particular, has a larger mechanical advantage than the quadriceps during periods of enhanced knee flexion (29), forcing the quadriceps to generate more compensatory force.

It has been proposed that prophylactic knee bracing facilitates safer movement mechanics by promoting an enhanced perception of joint stability (30). The subjective ratings support this notion, as participants perceived that the knee brace significantly improved knee joint stability. This investigation is the first to calculate lower extremity muscle kinetics whilst using prophylactic knee bracing during cycling. Active muscle stiffness promotes overall knee joint stability, and is proportionate to the extent of muscular activation and force production (31). Williams et al., (32) propose that joint mechanoreceptors contribute to joint stability by continually modulating muscle stiffness. As knee bracing enhanced subjective joint stability, we propose that joint mechanoreceptors detected this perceived change, allowing muscle forces to be proportionally reduced in the quadriceps and biceps femoris muscles in response to the presence of the brace.

Knee bracing also statistically reduced coronal plane maximum knee ROM. This concurs with those of Theobald et al., (16), who revealed that prophylactic bracing attenuated coronal plane ROM during cycling. This may be important, as retrospective analyses (33-35) have shown coronal plane knee kinematics to be enhanced in cyclists with patellofemoral pain. Therefore, this observation may provide further evidence to support the potential for prophylactic knee bracing to attenuate the risk from the biomechanical parameters linked to the aetiology of patellofemoral pain in cyclists. Theobald et al., (16) found that the brace examined in their study was too uncomfortable to be practically viable for adoption into practice. This observation does not agree with the subjective ratings provided during the current investigation, as although the Chi-Squared test was insignificant, 20 of the 24 participants rated the brace as either more comfortable or no-change. This indicates that discomfort may not be a significant barrier to the knee brace examined the current investigation being adopted clinically. The lack of alignment between studies is likely due to the differences in mechanical characteristics between the two experimental braces, as Theobald et al., (16) investigated a more structured device than that examined in the current study.

In conclusion, the current investigation adds to the current literature by providing a comparative examination of the effects of prophylactic knee bracing on cycling biomechanics during the pedal cycle using a musculoskeletal simulation approach. Importantly, the integral of patellofemoral stress during the pedal cycle and the maximum coronal plane knee ROM were significantly reduced in the brace condition. Furthermore, it was also revealed that that knee bracing significantly enhanced perceived knee joint stability compared to the no-brace condition. The findings from the current investigation therefore indicate that prophylactic knee bracing may have the potential to attenuate the biomechanical parameters linked to the aetiology of patellofemoral pain in cyclists. Future, longitudinal analyses are required to confirm the efficacy of prophylactic knee braces for the attenuation of patellofemoral pain symptoms in cyclists.

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