

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

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17
18 **Keywords:** Cycling; patellofemoral pain; knee brace; biomechanics; musculoskeletal
19 modeling.

20

21 **Abstract**

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

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

30 **RESULTS:** The results showed that the integral of the patellofemoral joint stress was
31 significantly lower in the brace condition (male: 70RPM=8.89, 80RPM=9.76, &
32 90RPM=12.30 KPa/kg·s and female: 70RPM=11.59, 80RPM=13.07 & 90RPM=14.14
33 KPa/kg·s) compared to no-brace (male: 70RPM=10.23, 80RPM=10.96 & 90RPM=13.20 and
34 female: 70RPM=12.43, 80RPM=14.04 & 90RPM=15.45 KPa/kg·s). In addition, it was also
35 revealed that participants rated that the knee brace significantly improved perceived knee joint
36 stability.

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

42

43 **Introduction**

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

49

50 Patellofemoral pain is the most frequently experienced musculoskeletal condition, affecting
51 36% of all cyclists and accounting for more than 57 % of all time-loss pathologies (4, 5).
52 Patellofemoral pain is so prevalent in cycling that it has been termed 'cyclist's knee' (6) and
53 the long term forecast for patients is poor, as many later present with radiographic
54 patellofemoral joint osteoarthritis (7). Elevated patellofemoral joint stress is the biomechanical
55 mechanism linked most strongly to the aetiological of patellofemoral pain (8), and although,
56 musculoskeletal modeling approaches exist to estimate patellofemoral joint loading (9, 10),
57 they require inverse dynamics as input parameters into the musculoskeletal algorithm. Joint
58 torques are not representative of localized joint loading, as Herzog et al., (11) showed that
59 muscles are the primary contributors to lower extremity joint kinetics. Recent advances in
60 musculoskeletal simulation software and associated models including the patellofemoral joint
61 (12) have been developed, which allow skeletal muscle force distributions to be simulated
62 during movement and utilized as input parameters for the quantification of lower extremity
63 joint loading. To date, there has been only limited utilization of musculoskeletal simulation for
64 cycling specific analyses.

65

66 Given the high incidence of patellofemoral pain in athletic and active populations, a range of
67 conservative prophylactic and treatment modalities have been explored in biomechanical and

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

81

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

88

89 **Methods**

90 *Participants*

91 Twenty-four recreational cyclists (12 male and 12 female), volunteered to take part in this
92 study. All had at least 2 years of road cycling experience and were from lower extremity
93 musculoskeletal pathology at the time of data collection. The mean characteristics of the
94 participants were; (males) age 28.14 ± 6.31 years, height 1.77 ± 0.07 m and body mass 79.04
95 ± 9.25 kg and (females) age 26.71 ± 5.65 years, height 1.64 ± 0.06 m and body mass $62.56 \pm$
96 7.33 kg. To be eligible for participation, cyclists were required to have at least 2 years of road
97 cycling experience. In addition, they were required to be free from musculoskeletal pathology
98 at the time of data collection, with no previous knee joint surgical intervention. The procedure
99 utilized for this investigation was approved by the University of Central Lancashire, Science,
100 Technology, Engineering and Mathematics, ethical committee (Ref: 644) and all participants
101 provided written informed consent

102

103 *Knee brace*

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

108

109 *Procedure*

110 Participants rode a stationary ergometer SRM 'Indoor Trainer' (SRM, Schoberer, Germany)
111 for 6 minutes at fixed cadences of 70, 80 and 90 RPM in both brace and no-brace conditions.
112 The experimental conditions were completed in a counterbalanced order and a standardized
113 rest period of 5 minutes was allowed between trials. The bicycle set-up was conducted in

114 accordance with previous recommendations (17), and maintained between each condition. The
115 cycling shoes (Northwave Sonic 2 Plus Road), pedals (Look Keo Classic 2, Look, Cedex,
116 France), cleats (Look Keo Grip, 4.5° float, Look, Cedex, France), chain ring (SRM power,
117 SRM, Schoberer, Germany) and crank (SRM power, SRM, Schoberer, Germany) were also
118 maintained across all trials, and positioned in accordance with previous recommendations (18).
119 Participants were given continuous visual feedback of their cadence, which was visible via the
120 SRM head unit (Powercontrol V, SRM, Schoberer, Germany).

121

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

138

139 In addition to the biomechanical movement information, the effects of the experimental brace
140 on knee joint proprioception were also examined using a cycling specific joint position sense
141 test. This was conducted, in accordance with the procedure of Drouin et al., (29), whereby
142 participants were assessed on their ability to reproduce a target knee flexion angle whilst sat
143 on the cycle ergometer. To accomplish this, participants were asked to slowly turn the pedal to
144 90 ° from the point of top dead centre, which was verified using a handheld goniometer by the
145 same researcher throughout data collection. Participants then held this position for 15 seconds
146 during which time the ‘criterion’ knee flexion position was captured using the motion analysis
147 system. Following this, participants were asked to pedal at a fixed cadence of 60 RPM for 60
148 seconds, after which they reproduced the target position as accurately as possible but without
149 guidance via the goniometer. Again, this position was held for a period of 15 seconds and the
150 knee flexion angle during the ‘replication’ trial was also collected using the motion analysis
151 system. This above process was conducted on three occasions in both the brace and no-brace
152 conditions in a counterbalanced order. The absolute difference in degrees calculated between
153 the criterion and replication trials was averaged over the three trials to provide angular error
154 values in both brace and no-brace conditions, which were extracted for statistical analysis.

155

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

161

162 *Processing*

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

166

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

175

176 Data from the five pedal cycles in each condition were then exported from Visual 3D into
177 OpenSim 3.3 software (Simtk.org). A validated musculoskeletal model with 12 segments, 19
178 degrees of freedom and 92 musculotendon actuators (12) was used to quantify patellofemoral
179 joint forces. The model was firstly scaled for each participant to account for the
180 anthropometrics of each rider. We firstly performed a residual reduction algorithm (RRA)
181 within OpenSim; in order to reduce the residual forces and moments (22). As muscle forces
182 are the main determinant of joint compressive forces (11), muscle kinetics were quantified
183 using a static optimization process in accordance with Steele et al., (23). Following this
184 patellofemoral, joint forces were calculated using the joint reaction analyses function using the
185 muscle forces generated from the static optimization process as inputs. Finally, patellofemoral
186 joint stress was quantified by dividing the patellofemoral force by the patellofemoral contact

187 area. Patellofemoral contact area were obtained by fitting a 2nd order polynomial curve to the
188 sex specific data of Besier et al., (24), who estimated patellofemoral contact areas as a function
189 of the knee flexion angle using MRI.

190

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

203

204 *Statistical analyses*

205 Descriptive statistics of means and standard deviations were obtained for each outcome
206 measure. Shapiro-Wilk tests were used to screen the data for normality. Differences in knee
207 proprioception with and without the presence of the brace were examined using a 2 (BRACE)
208 x 2 (GENDER) mixed ANOVA. Differences in biomechanical parameters were examined
209 using 2 (BRACE) x 3 (WORKLOAD) x 2 (GENDER) mixed ANOVA's. In the event of a

210 significant main effect, pairwise comparisons were performed and any significant interactions
211 were explored using simple main effects. In addition, the subjective ratings in relation to the
212 stability and comfort of the knee sleeve were examined using Chi-Squared (χ^2) tests. Statistical
213 significance was accepted at the $P \leq 0.05$ level. Effect sizes for all significant findings were
214 calculated using partial Eta² (η^2). All statistical actions were conducted using SPSS v24.0
215 (SPSS Inc, Chicago, USA).

216

217 **Results**

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

220

221 *Patellofemoral joint kinetics and contact area*

222 For peak patellofemoral force, a significant main effect of WORKLOAD was observed
223 ($P < 0.05$, $\eta^2 = 0.18$). Post-hoc pairwise comparisons showed that peak force was statistically
224 larger in the 90 RPM condition compared to 70 RPM ($P = 0.02$) (*Table 1 & 2*). In addition, for
225 peak patellofemoral stress, a significant main effect of WORKLOAD was shown ($P < 0.05$, η^2
226 $= 0.17$). Post-hoc pairwise comparisons showed that peak force was statistically larger in the
227 90 RPM condition compared to 70 RPM ($P = 0.03$) (*Table 1 & 2*).

228

229 For the integral of the patellofemoral joint force, significant main effects of both WORKLOAD
230 ($P < 0.05$, $\eta^2 = 0.14$) and BRACE ($P < 0.05$, $\eta^2 = 0.28$) were noted. Post-hoc pairwise
231 comparisons for WORKLOAD showed that the patellofemoral force integral was statistically

232 larger in the 90 (P=0.04) and 80 RPM (P=0.03) conditions compared to 70 RPM. For BRACE
233 it was shown that the integral of the patellofemoral joint force was statistically larger in the no-
234 brace condition (P=0.008) (*Table 1 & 2*). In addition, for the integral of the patellofemoral joint
235 stress, a significant main effect of for BRACE (P<0.05, $p\eta^2 = 0.27$) was noted, with the
236 patellofemoral integral stress being statistically larger in the no-brace condition (P=0.009)
237 (*Table 1 & 2*).

238

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

240

241 @@@TABLE 1 NEAR HERE@@@

242 @@@TABLE 2 NEAR HERE@@@

243

244 *Muscle kinetics*

245 For the peak rectus femoris force a significant main effect of WORKLOAD (P<0.05, $p\eta^2 =$
246 0.31) was found. Post-hoc pairwise comparisons showed that peak force was statistically larger
247 in the 90 RPM compared to the 70 (P=0.002) and 80 RPM (P=0.03) conditions and that 80
248 RPM was larger than 70 RPM (P=0.0004) (*Table 3 & 4*). For the integral of the rectus femoris
249 force a significant BRACE main effect was found (P<0.05, $p\eta^2 = 0.23$), with the integral force
250 being statistically larger in the no-brace condition (P=0.02) (*Table 3 & 4*).

251

252 For the peak vastus lateralis force, significant main effects of WORKLOAD (P<0.05, $p\eta^2 =$
253 0.18) and BRACE (P<0.05, $p\eta^2 = 0.21$) were found. Post-hoc pairwise comparisons for

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

257

258 For the peak vastus medialis force, significant main effects of WORKLOAD (P<0.05, $\eta^2 =$
259 0.17) and BRACE (P<0.05, $\eta^2 = 0.24$) were found. Post-hoc pairwise comparisons for
260 WORKLOAD showed that peak force was statistically larger in the 90 RPM (P=0.03)
261 condition than 70 RPM. For BRACE the peak force was statistically larger in the no-brace
262 condition (P=0.02) (*Table 3 & 4*). For the integral of the vastus medialis force a significant
263 BRACE main effect was found (P<0.05, $\eta^2 = 0.17$), with the integral force being statistically
264 larger in the no-brace condition (P=0.04) (*Table 3 & 4*).

265

266 For the peak vastus intermedius force, significant main effects of WORKLOAD (P<0.05, η^2
267 = 0.17) and BRACE (P<0.05, $\eta^2 = 0.27$) were found. Post-hoc pairwise comparisons for
268 WORKLOAD showed that peak force was statistically larger in the 90 RPM (P=0.03)
269 condition than 70 RPM. For BRACE the peak force was statistically larger in the no-brace
270 condition (P=0.009) (*Table 3 & 4*). For the integral of the vastus intermedius force a significant
271 BRACE main effect was found (P<0.05, $\eta^2 = 0.17$), with the integral force being statistically
272 larger in the no-brace condition (P=0.04) (*Table 3 & 4*).

273

274 For the peak biceps femoris long head force, significant main effects of WORKLOAD (P<0.05,
275 $\eta^2 = 0.29$) and BRACE (P<0.05, $\eta^2 = 0.34$) were found. Post-hoc pairwise comparisons for
276 WORKLOAD showed that peak force was statistically larger in the 80 (P=0.001) and 90 RPM

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

281

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

286

287 For the peak semimembranosus force, a significant main effect of WORKLOAD (P<0.05, η^2
288 = 0.18) was found. Post-hoc pairwise comparisons showed that peak force was statistically
289 larger in the 90 (P=0.03) and 80 RPM (P=0.02) conditions compared to 70 RPM (*Table 3 &*
290 *4*).

291

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

295

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

297

298

@@@TABLE 3 NEAR HERE@@@

299

@@@TABLE 4 NEAR HERE@@@

300

301 *Three-dimensional kinematics*

302 In the sagittal plane, a significant main effect of WORKLOAD ($P < 0.05$, $\eta^2 = 0.20$) was found.

303 Post-hoc pairwise comparisons showed that the sagittal plane maximum knee range of motion

304 (ROM) was statistically larger in the 90 RPM compared to the 70 ($P = 0.02$) and 80 RPM

305 ($P = 0.006$) conditions (*Table 5 & 6*).

306

307 In the coronal plane, significant main effects of WORKLOAD ($P < 0.05$, $\eta^2 = 0.22$) and

308 BRACE ($P < 0.05$, $\eta^2 = 0.24$) were found. Post-hoc pairwise comparisons showed that the

309 coronal plane maximum knee ROM was statistically larger in the 90 RPM compared to the 70

310 ($P = 0.02$) and 80 RPM ($P = 0.02$) conditions (*Table 5 & 6*). For BRACE maximum coronal knee

311 ROM was statistically larger in the no-brace condition ($P = 0.02$) (*Table 5 & 6*).

312

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

314

315 **@@@TABLE 5 NEAR HERE@@@**

316 **@@@TABLE 6 NEAR HERE@@@**

317

318 *Knee proprioception*

319 No significant differences ($P>0.05$) in knee proprioception were shown. In the no-brace
320 condition, a mean error of $4.70 \pm 2.59^\circ$ was found for males and $6.90 \pm 4.05^\circ$ shown for
321 females. In the brace condition, a mean error of $3.74 \pm 2.58^\circ$ was found for males had and 6.34
322 $\pm 3.60^\circ$ shown for females.

323

324 *Subjective preferences*

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

329

330 **Discussion**

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

341

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

353

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

364

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

376

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

390 between the two experimental braces, as Theobald et al., (16) investigated a more structured
391 device than that examined in the current study.

392

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

404

405 **References**

- 406 1. Clarsen B, Krosshaug T, Bahr R (2010). Overuse injuries in professional road cyclists.
407 Am J Sports Med 38: 2494–2501. doi:10.1177/0363546510376816
- 408 2. Hollingworth MA, Harper AJ, Hamer M. (2015). Risk factors for cycling accident
409 related injury: the UK Cycling for Health Survey. J Transp Health 2: 189-194.
410 doi:10.1016/j.jth.2015.01.001

- 411 3. Rissel C, Passmore E, Mason C, Merom D. (2013). Two pilot studies of the effect of
412 bicycling on balance and leg strength among older adults. *J Environ Public Health* 6:
413 24-30. [doi:10.1155/2013/686412](https://doi.org/10.1155/2013/686412)
- 414 4. Wanich T, Hodgkins C, Columbier JA, Muraski E, Kennedy JG. (2007). Cycling
415 injuries of the lower extremity. *J Am Acad Orthop Surg* 15: 748-756. [doi:](https://doi.org/10.5435/00124635-200712000-00008)
416 [10.5435/00124635-200712000-00008](https://doi.org/10.5435/00124635-200712000-00008)
- 417 5. Dieter BP, McGowan CP, Stoll SK, Vella CA (2014). Muscle activation patterns and
418 patellofemoral pain in cyclists. *Med Sci Sports Exerc* 46: 753-761. [doi:](https://doi.org/10.1249/MSS.0000000000000153)
419 [10.1249/MSS.0000000000000153](https://doi.org/10.1249/MSS.0000000000000153)
- 420 6. Callaghan, M.J. (2005). Lower body problems and injury in cycling. *J Bodyw Mov*
421 *Ther* 9: 226-236. [doi:10.1016/j.jbmt.2005.01.007](https://doi.org/10.1016/j.jbmt.2005.01.007)
- 422 7. Thomas MJ, Wood L, Selfe J, Peat G. (2010). Anterior knee pain in younger adults as
423 a precursor to subsequent patellofemoral osteoarthritis: a systematic review. *BMC*
424 *Musc Dis* 11: 201-205. [doi: 10.1186/1471-2474-11-201](https://doi.org/10.1186/1471-2474-11-201)
- 425 8. Farrokhi S, Keyak JH, Powers CM. (2011). Individuals with patellofemoral pain exhibit
426 greater patellofemoral joint stress: a finite element analysis study. *Osteoarthr Cartil* 19:
427 287-294. [doi: 10.1016/j.joca.2010.12.001](https://doi.org/10.1016/j.joca.2010.12.001)
- 428 9. van Eijden TM, Kouwenhoven E, Verburg J, Weijs WA (1986). A mathematical model
429 of the patellofemoral joint. *J Biomech* 19: 219–229. [doi: 10.1016/0021-9290\(86\)90154-](https://doi.org/10.1016/0021-9290(86)90154-5)
430 [5](https://doi.org/10.1016/0021-9290(86)90154-5)
- 431 10. Elias JJ, Wilson DR, Adamson R, Cosgarea AJ (2004). Evaluation of a computational
432 model used to predict the patellofemoral contact pressure distribution. *J Biomech* 37:
433 295-302. [doi:10.1016/S0021-9290\(03\)00306-3](https://doi.org/10.1016/S0021-9290(03)00306-3)
- 434 11. Herzog W, Longino D, Clark A (2003). The role of muscles in joint adaptation and
435 degeneration. *Langenbecks Arch Surg* 388: 305-315. [doi:10.1007/s00423-003-0402-6](https://doi.org/10.1007/s00423-003-0402-6)

- 436 12. Lerner ZF, DeMers MS, Delp SL, Browning RC. (2015). How tibiofemoral alignment
437 and contact locations affect predictions of medial and lateral tibiofemoral contact
438 forces. *J Biomech* 48: 644-650. doi: [10.1016/j.jbiomech.2014.12.049](https://doi.org/10.1016/j.jbiomech.2014.12.049)
- 439 13. Martin TJ. (2001). Technical report: knee brace use in the young athlete. *Pediatrics* 108:
440 503-507.
- 441 14. Warden SJ, Hinman RS, Watson MA, Avin KG, Bialocerowski AE, Crossley KM.
442 (2008). Patellar taping and bracing for the treatment of chronic knee pain: A systematic
443 review and meta-analysis. *Arthritis Care Res* 59: 73-83. doi: [10.1002/art.23242](https://doi.org/10.1002/art.23242).
- 444 15. Paluska SA, McKeag DB. (2000). Knee braces: current evidence and clinical
445 recommendations for their use. *American Family Physician* 61: 411-418. doi:
446 [10.1097/PHM.0b013e318114e416](https://doi.org/10.1097/PHM.0b013e318114e416)
- 447 16. Theobald G, Parry S, Richards J, Thewlis D, Tootill I, Selfe J. (2012). Biomechanical
448 effects of different treatment modalities used in knee pain during cycling.
449 *Physiotherapy Practice and Research* 33: 16-21.
- 450 17. Sinclair J, Stainton P, Sant B. (2018). The effects of conventional and oval chainrings
451 on patellofemoral loading during road cycling: an exploration using musculoskeletal
452 simulation. *Sport Sci Health* 14: 61-70. doi: [10.1007/s1133](https://doi.org/10.1007/s1133)
- 453 18. Menard, M., Domalain, M., Decatoire, A., Lacouture, P. (2016). Influence of saddle
454 setback on pedalling technique effectiveness in cycling. *Sport Biomech* 15: 462-472.
455 doi: [10.1080/14763141.2016.1176244](https://doi.org/10.1080/14763141.2016.1176244)
- 456 19. Drouin JM, Houglum PA, Perrin DH, Gansneder BM. (2003). Weight bearing and non-
457 weight-bearing knee joint reposition sense are not related to functional performance. *J*
458 *Sport Rehab* 12: 54-66. doi: [10.1123/jsr.12.1.54](https://doi.org/10.1123/jsr.12.1.54)

- 459 20. Sinclair J, Vincent H, Richards D. (2017). Effects of prophylactic knee bracing on knee
460 joint kinetics and kinematics during netball specific movements. *Phys Ther Sport* 23:
461 93-98. doi: [10.1016/j.ptsp.2016.08.005](https://doi.org/10.1016/j.ptsp.2016.08.005)
- 462 21. Sinclair J, Hebron J, Atkins S, Hurst H, Taylor PJ. (2014). The influence of 3D
463 kinematic and electromyographical parameters on cycling economy. *Acta Bioeng
464 Biomech* 16: 91-97. doi: [10.5277/ABB-00049-2014-02](https://doi.org/10.5277/ABB-00049-2014-02)
- 465 22. Delp SL, Anderson FC, Arnold AS, Loan P, Habib A, John CT, Thelen DG (2007).
466 OpenSim: open-source software to create and analyze dynamic simulations of
467 movement. *IEEE Trans Biomed Eng* 54: 1940-1950. doi: [10.1109/TBME.2007.901024](https://doi.org/10.1109/TBME.2007.901024)
- 468 23. Steele KM, DeMers MS, Schwartz MH, Delp SL. (2012). Compressive tibiofemoral
469 force during crouch gait. *Gait Posture*, 35: 556-560. doi:
470 [10.1016/j.gaitpost.2011.11.023](https://doi.org/10.1016/j.gaitpost.2011.11.023)
- 471 24. Besier TF, Draper CE, Gold GE, Beaupre GS, Delp SL. Patellofemoral joint contact
472 area increases with knee flexion and weight-bearing. *J Orthop Res* 2005; 23: 345–350.
473 doi: [10.1016/j.orthres.2004.08.003](https://doi.org/10.1016/j.orthres.2004.08.003)
- 474 25. Besier TF, Fredericson M, Gold GE, Beaupré GS, Delp SL. (2009). Knee muscle forces
475 during walking and running in patellofemoral pain patients and pain-free controls. *J
476 Biomech* 42: 898-905. doi: [10.1016/j.jbiomech.2009.01.032](https://doi.org/10.1016/j.jbiomech.2009.01.032)
- 477 26. Elias JJ, Kirkpatrick MS, Saranathan A, Mani S, Smith LG, Tanaka MJ. (2011).
478 Hamstrings loading contributes to lateral patellofemoral malalignment and elevated
479 cartilage pressures: an in vitro study. *Clini Biomech* 26: 841-846. doi:
480 [10.1016/j.clinbiomech.2011.03.016](https://doi.org/10.1016/j.clinbiomech.2011.03.016)
- 481 27. Whyte EF, Moran, K, Shortt, CP, Marshall B. (2010). The influence of reduced
482 hamstring length on patellofemoral joint stress during squatting in healthy male adults.
483 *Gait Posture* 31: 47-51. doi: [10.1016/j.gaitpost.2009.08.243](https://doi.org/10.1016/j.gaitpost.2009.08.243)

- 484 28. Li G, DeFrate LE, Zayontz S, Park SE, Gill T J. (2004). The effect of tibiofemoral joint
485 kinematics on patellofemoral contact pressures under simulated muscle loads. Journal
486 Orthopaedic Res 22: 801-806. [doi:10.1016/j.orthres.2003.11.011](https://doi.org/10.1016/j.orthres.2003.11.011)
- 487 29. Wretenberg P, Németh G, Lamontagne M, Lundin B. (1996). Passive knee muscle
488 moment arms measured in vivo with MRI. Clin Biomech 11: 439-446. [doi:](https://doi.org/10.1016/S0268-0033(96)00030-7)
489 [10.1016/S0268-0033\(96\)00030-7](https://doi.org/10.1016/S0268-0033(96)00030-7)
- 490 30. Herrington L, Simmonds C, Hatcher J. (2005). The effect of a neoprene sleeve on knee
491 joint position sense. Res Sport Med 13: 37-46. [doi: 10.1080/15438620590922077](https://doi.org/10.1080/15438620590922077)
- 492 31. Julian FJ, Morgan DL. (1981). Variation of muscle stiffness with tension during tension
493 transients and constant velocity shortening in the frog. J. Physiol 319: 193–203. [doi:](https://doi.org/10.1113/jphysiol.1981.sp013901)
494 [10.1113/jphysiol.1981.sp013901](https://doi.org/10.1113/jphysiol.1981.sp013901)
- 495 32. Williams GN, Chmielewski T, Rudolph KS, Buchanan TS, Snyder-Mackler L. (2001).
496 Dynamic knee stability: current theory and implications for clinicians and scientists. J
497 Orthop Sports Phys Ther 31: 546-566. [doi: 10.2519/jospt.2001.31.10.546](https://doi.org/10.2519/jospt.2001.31.10.546)
- 498 33. Hannaford DR, Moran GT, Hlavac HF. (1986). Video analysis and treatment of overuse
499 knee injury in cycling: a limited clinical study. Clin Podiatr Med Surg 3: 671–678.
- 500 34. Bailey M, Messenger N. (1995). Coronal plane kinematics of cycling and their
501 relationship to injury. J Sport Sci 14: 3–4.
- 502 35. Bailey M, Maillardet F, Messenger N. (2003). Kinematics of cycling in relation to
503 anterior knee pain and patellar tendinitis. J Sport Sci 21: 649-657. [doi:](https://doi.org/10.1080/0264041031000102015)
504 [10.1080/0264041031000102015](https://doi.org/10.1080/0264041031000102015)