

1 **Effects of a prophylactic knee sleeve on anterior cruciate ligament and lower extremity**

2 **biomechanics: an examination using musculoskeletal simulation.**

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20
21 **Abstract**

22 The current study aimed using a two-experiment musculoskeletal simulation-based approach,
23 measuring ACL biomechanics, knee joint kinematics and lower extremity joint loading to
24 examine the effects of both a prophylactic knee sleeve on 1. a sport specific change of direction
25 movement in female footballers and 2. a single leg landing in male footballers. **Experiment 1**

26 examined 12 female university first team level footballers (age 20.2 ± 1.34 years, height
27 1.61 ± 0.06 m, body mass 57.2 ± 5.6 kg) undertaking a 45° cutting movement in sleeve and no-
28 sleeve conditions. Experiment 2 examined 10 male university first team level footballers (age
29 21.1 ± 1.13 years, height 1.77 ± 0.1 m, body mass 71.9 ± 8.6 kg) undertaking a single leg drop
30 jump landing in sleeve and no-sleeve conditions. In each experiment, data was collected in a
31 biomechanics laboratory and three-dimensional motion capture and ground reaction force
32 information was collected. Three-dimensional kinematics, three-dimensional knee kinetics and
33 ACL ligament forces/ strains were measured using musculoskeletal simulation, and
34 participants were also asked to subjectively rate the knee sleeve in terms of both comfort and
35 stability. Experiment 1 showed that the sleeve condition was associated with greater ACL strain
36 (sleeve = 13.57% and no-sleeve = 10.26%) and forces (sleeve = 1.19BW and no-sleeve =
37 0.94BW). In addition, the brace condition also enhanced lateral compressive tibiofemoral
38 (sleeve = 4.70BW and no-sleeve = 4.20BW) and total compressive tibiofemoral force (sleeve
39 = 11.73BW and no-sleeve = 11.08BW). Finally, for the subjective ratings, participants
40 indicated that the knee sleeve significantly improved perceived comfort and stability.
41 Experiment 2 did not reveal any statistical differences between knee sleeve and no-sleeve
42 conditions, nor any effects of the knee sleeve on subjective ratings of comfort or stability.
43 Therefore, the findings from the current investigation suggest that the prophylactic knee sleeve
44 examined in the current investigation does not appear to reduce the biomechanical parameters
45 linked to the aetiology of knee pathologies in male/ female footballers.

46

47 **Introduction**

48 Football is regarded as the most popular sport in terms of audience and participants, with more
49 than 200,000 professional and over 240 million amateur players globally ¹. Football like most
50 other team sports is characterized by intermittent deceleration and landing activities requiring

51 rapid and agile change of direction movements ². As both a competitive and recreational
52 activity, football is associated with a plethora of physical benefits including enhanced
53 cardiovascular, mental and bone health ³. However, football is also connected with a relatively
54 high incidence of injury ⁴, which has been shown to exert a significant burden on
55 socioeconomic and healthcare systems ⁵. Epidemiological investigations in professional
56 players have shown injury rates of 8.0 per 1000 h and an average of 2.0 injuries per season ⁶
57 and 38.56 per 1000 h, at a rate of 0.85 time-loss injuries per match in recreational players ⁷.

58

59 One of the most commonly injured musculoskeletal structures in football is the knee ^{6,7}, and
60 the anterior cruciate ligament (ACL) is the most frequently injured knee ligament ⁸. The ACL
61 itself is vital for the provision of knee stability during the dynamic activities associated with
62 football ⁹. With its unique functional properties, attachment points and complex anatomy, the
63 ACL is highly effective in restraining both excessive anterior tibial translation and coronal/
64 transverse plane knee motions ¹⁰. ACL injuries in football players are predominantly, non-
65 contact in nature, in that the ligament becomes injured without physical contact between
66 players ¹¹.

67

68 Physiologically, ACL injuries occur when the ligament experiences excessive tensile forces
69 and strains ¹². As the ACL serves primarily to resist anteriorly directed tibial translation in
70 addition to knee valgus and internal/ external rotation movements; in vivo and in vitro
71 investigations have shown that it experiences both load and strain during activities that involve
72 these mechanisms ¹³. Aetiological investigations support this, in that the ACL is most
73 commonly disrupted in the period immediately following foot contact, in athletic tasks
74 involving sudden decelerations, landings and cutting manoeuvres ¹⁴. Injury to the ACL is

75 extremely serious in competitive players, and typically leads to long term absence from football
76 ¹⁵. ACL pathologies typically require reconstructive intervention using auto/allografts in order
77 to provide sufficient stability to the injured knee to allow return to training/ competitive
78 activities ^{16, 17}. Silvers & Mandelbaum ¹⁸ showed that over 250,000 ACL reconstruction
79 interventions are undertaken each year in the US alone with average allocated costs exceeding
80 \$2 billion.

81

82 Importantly, the ACL can be associated with poor healing capacity, and the risk of a second
83 injury is as high as 30% in the ipsilateral knee and 11% in the contralateral side ^{19, 20}. Even
84 after full recovery, ACL injuries frequently lead to chronic knee pain, and athletes who
85 experience an ACL pathology are up to ten times more susceptible to early-onset degenerative
86 knee osteoarthritis ²¹, leading not only to a decline in athletic participation but also enduring
87 disability in later life ²². Radiographic knee osteoarthritis significantly reduces health-related
88 quality of life, and degenerative joint disease secondary to ACL injury imposes further
89 economic burden ²³. Similarly, it has been demonstrated that psychological as well as physical
90 wellbeing is negatively affected, and ACL injuries have been associated with anxiety, self-
91 esteem, pain response, depression, and feelings of decreased athletic identity ²⁴. Importantly,
92 previous analyses have shown that many footballers fail to return to their previous levels of
93 athletic function, as statistically significant performance decrements have been observed in
94 relation to non-injured controls ²⁵. Concerningly, both Roos et al., ²⁶ and Walden et al., ¹⁵
95 demonstrated that only 30-35% of competitive footballers remained active 3 years after
96 suffering an ACL injury.

97

98 Because of the high incidence of ACL injuries in football players ¹⁵ and the poor-long term
99 prognosis following injury, prophylactic interventions are therefore a key clinical priority ²⁷.
100 Knee braces are external devices constructed in order to improve three-dimensional knee joint
101 dynamic alignment ²⁸ and range from semi-rigid devices incorporating uni or polyaxial hinges
102 to more compliant sleeves designed simply to provide compression and enhance proprioception
103 ²⁹. Knee braces represent a conservative and relatively low-cost external apparatus that are
104 minimally invasive/ restrictive such that they can be worn during high-intensity sports
105 maneuvers ²⁸. Prophylactic knee braces have been shown to reduce transverse plane knee range
106 of motion during run, cut and vertical jump movements in netball players ²⁸, peak knee
107 adduction moment during a badminton lunge ³⁰ and patellar tendon loading in run, cut and
108 single leg hop movements in female athletes ³¹. Furthermore, Sinclair et al. ³² showed using an
109 inverse dynamics-based method of quantifying ligament loading, that ACL load rates were
110 significantly reduced during single leg hop landings and cut movements.

111

112 However, the efficacy of any intervention modality depends on a sound comprehension of the
113 underlying causative mechanisms of the associated condition. Inverse dynamics represent only
114 global indices of joint loading, and therefore, are not truly representative of localized loading
115 experienced by the joint structures ³³. Herzog et al. ³⁴ showed that muscles are the primary
116 contributors to the forces experienced by the lower extremity joint structures. Specifically, the
117 complex role of muscles in controlling knee ligament loading during human movement has
118 received insufficient attention within the literature, owing to difficulties in calculating muscle
119 kinetics and modelling knee joint ligamentous structures ²⁷. To date, there has yet to be any
120 investigation which has examined the effects of prophylactic knee bracing on ligament load
121 and strain parameters linked to the aetiology of ACL using a muscle driven approach to

122 quantify knee mechanics. This is principally due to the inability to non-invasively quantify
123 ACL loads and strains during high-risk sports movements³⁵.

124

125 Recent, advances in musculoskeletal simulation software alongside enhancements in
126 simulation model algorithmic complexity, mean that quantitative indices of ACL kinetics and
127 strains are now attainable alongside more traditional simulation parameters of joint and muscle
128 forces³⁶. To date however, this more advanced modelling approach has not yet been utilized
129 to explore the effects of prophylactic knee sleeves on ACL loading and strain during high-risk
130 sports specific football movements. Similarly, whilst the effects of prophylactic knee sleeve
131 have been examined previously, they have focused only on indices of knee joint loading/
132 kinematics. Knee sleeves are likely to mediate both kinetic and kinematic alterations at more
133 than one body segment and thus at more than one joint; and potential positive alterations at the
134 knee joint mediated via the sleeve, may cause concurrent effects at other lower extremity joints.
135 Therefore, a more comprehensive approach also examining hip and ankle joint loading in
136 addition to knee joint kinetics would be of both practical and clinical relevance.

137

138 To summarize, there is currently no scientific investigation that has explored the effects of
139 prophylactic knee bracing on collective indices of ACL loading/ strains alongside lower
140 extremity joint loading using musculoskeletal simulation in football players. Therefore, the
141 aims of the current study were, using a two-experiment musculoskeletal simulation-based
142 approach (whilst measuring ACL biomechanics, knee joint kinematics and lower extremity
143 joint loading) to examine the effects of both a prophylactic knee sleeve on 1. a sport specific
144 cutting movement in female university level footballers and 2. a single leg landing in male
145 university footballers. A study of this nature may provide further insight into the

146 comprehensive biomechanical effects of prophylactic knee sleeve designed to reduce the risk
147 from knee pathologies in football players.

148

149 **Methods**

150 For both investigations, participants provided written informed consent and ethical approval
151 was obtained from the University of Central Lancashire, in accordance with the principles
152 documented in the Declaration of Helsinki. All participants were free from lower extremity
153 musculoskeletal pathology at the time of data collection and had not undergone surgical
154 intervention at the knee joint.

155

156 *Knee sleeve*

157 A single nylon/silicone knee sleeve (Figure 1) was utilized in this investigation, (Kuangmi 1
158 PC compression knee sleeve), was used in this study which came in three different sizes;
159 small, medium and large to accommodate all participants and was worn on the dominant
160 (right) limb in all participants. In accordance with Sinclair et al.,²⁸, at the end of data
161 collection participants were asked to subjectively rate the knee sleeve in relation to
162 performing the movements without the sleeve in terms of stability and comfort. This was
163 accomplished using 3-point scales that ranged from 1 = increased comfort, 2 = no-change
164 and 3 = reduced comfort and 1 = increased stability, 2 = no change and 3 = increased stability.

165

166 **@@@FIGURE 1 NEAR HERE@@@**

167

168 *Experiment 1*

169 *Participants*

170 Twelve female (age 20.2 ± 1.34 years, height 1.61 ± 0.06 m, body mass 57.2 ± 5.6 kg and
171 BMI = 22.1 ± 3.0 kg/m²) university first team level footballers volunteered to take part in the
172 current investigation.

173

174 *Procedure*

175 Participants completed five trials of a 45° cut movement in both experimental conditions
176 (sleeve and no-sleeve). Data collection was undertaken in 22 m long biomechanics laboratory,
177 using an a-priori approach velocity of 4.0 ± 0.2 m/s striking the force platform with their right
178 (dominant) limb. Cut angles were measured from the centre of the force platform and the
179 corresponding line of movement was delineated using masking tape so that it was clearly
180 evident to participants (Figure 2). The stance phase of the cut movement was defined as the
181 duration over > 20 N of vertical force applied to the force platform.

182

183

@@@FIGURE 2 NEAR HERE@@@

184

185 The order in which participants performed in each knee sleeve condition was counterbalanced
186 i.e. participant 1 performed first in the knee sleeve condition followed by the no-sleeve
187 condition whereas participant 2 was examined first in the no-sleeve condition followed by the
188 knee sleeve and so on and so forth. To ensure consistency, each participant wore the same
189 footwear (Asics, Patriot 6). Kinematic information was obtained using an eight-camera wall
190 mounted motion analysis system (Qualisys Medical AB, Goteburg, Sweden) with a capture
191 frequency of 250 Hz. The camera system was arranged in an umbrella-based configuration and
192 covered an 8 m length and 6 m width (Figure 2). To measure ground reaction forces (GRF), an
193 embedded piezoelectric force platform (Kistler National Instruments, Model 9281CA)

194 operating at 1000 Hz was adopted. The GRF and kinematic information were synchronously
195 obtained using an analogue board and interfaced using Qualisys track manager.

196

197 To define the anatomical frames of the thorax, pelvis, thighs, shanks and feet, passive
198 retroreflective markers of 19mm diameter were placed at the C7, T12 and xiphoid process
199 landmarks and also positioned bilaterally onto the acromion process, iliac crest, anterior
200 superior iliac spine (ASIS), posterior superior iliac spine (PSIS), medial and lateral malleoli,
201 medial and lateral femoral epicondyles, greater trochanter, calcaneus, first metatarsal and fifth
202 metatarsal (Figure 3a). The hip, knee and ankle joint centre's were delineated according to
203 previously established guidelines³⁷⁻³⁹. Carbon-fibre tracking clusters comprising of four non-
204 linear retroreflective markers were positioned onto the thigh and shank segments. The foot
205 segments were tracked via the calcaneus, first and fifth metatarsal, the pelvic segment using
206 the PSIS and ASIS markers and the thorax via the T12, C7 and xiphoid markers. Static
207 calibration trials were obtained with the participant in the anatomical position in order for the
208 positions of the anatomical markers to be referenced in relation to the tracking clusters/markers,
209 following which those not required for dynamic data were removed. The Z (transverse) axis
210 was oriented vertically from the distal segment end to the proximal segment end. The Y
211 (coronal) axis was oriented in the segment from posterior to anterior. Finally, the X (sagittal)
212 axis orientation was determined using the right-hand rule and was oriented from medial to
213 lateral (Figure 3b).

214

215 **@@@FIGURE 3 NEAR HERE@@@**

216

217 Furthermore, the effects of the prophylactic sleeve on knee joint proprioception were
218 investigated via a weight-bearing knee joint position sense test. In accordance with the

219 procedure of Sinclair et al.²⁹, (with all of the above-mentioned retroreflective markers
220 remaining in place) participants stood in the centre of the motion capture system volume, on
221 one leg using the dominant limb. They then slowly squatted to a knee flexion angle of 30°,
222 which was verified using a handheld goniometer via same researcher throughout the testing
223 process. This position was held for a period of 15 s during which time the knee ‘criterion’ angle
224 was captured using the motion capture system (Figure 4ab). Following this, participants were
225 asked to return to a standing (i.e. with both feet on the floor) position for a further 15 s, and
226 then repeated the above process without guidance from the goniometer; a condition henceforth
227 named ‘unaided’. This position was again held for a period of 15 s and the unaided trial was
228 similarly collected using the motion analysis system. This above process was undertaken on
229 three occasions in both prophylactic sleeve and no-sleeve conditions using a counterbalanced
230 order, and in between each trial participants walked a fixed distance of 20 ft to eliminate
231 proprioceptive memory of the previous trial.

232

233 **@@@FIGURE 4 NEAR HERE@@@**

234

235 *Data Processing*

236 Dynamic and proprioception trials were digitized using Qualisys Track Manager (Qualisys
237 Medical AB, Goteburg, Sweden) in order to identify anatomical and tracking markers then
238 exported as C3D files to Visual 3D (C-Motion, Germantown, MD, USA). GRF data and marker
239 trajectories were smoothed with cut-off frequencies of 50 Hz at 12 Hz respectively, using a
240 low-pass Butterworth 4th order zero lag filter. Within Visual 3D knee joint angles were
241 quantified using an XYZ cardan sequence (where X is the sagittal plane; Y is the coronal plane
242 and is Z is the transverse plane).

243

244 For the proprioceptive data, the knee flexion angle during the criterion and unaided trials was
245 calculated. The absolute difference in the knee flexion angle in degrees, was calculated between
246 the criterion and unaided trials to provide an proprioception angular error value for both the
247 prophylactic knee sleeve and no-sleeve conditions (with a low value indicates greater knee
248 proprioception) and then extracted for statistical analysis. For dynamic trials obtained during
249 the 45° cut movements, these were linearly normalized to 100 % of the stance phase. Three-
250 dimensional angular kinematic measures from the stance phase that were extracted from the
251 knee joint in each of the angular planes of rotation were peak angle, peak angular velocity and
252 minimum angular velocity.

253

254 Dynamic data during the stance phase was exported from Visual 3D into OpenSim 3.3 software
255 (Simtk.org) using a custom pipeline that allowed the inverse kinematics to be exported to match
256 the degrees of freedom associated with the experimental model in OpenSim ²⁷. The standard
257 Gait2392 Opensim musculoskeletal model was adapted to include six degrees of freedom knee
258 joints and also an ACL bundles modelled in accordance with Sinclair et al., ²⁷ as non-linearly
259 elastic passive soft tissues based on the proximal (femur) and distal (tibia) insertion points of
260 Xu et al., ⁴⁰ (Figure 5ab). The model was further developed by incorporating a patella and the
261 tibiofemoral joint was separated into medial and lateral compartment locations which were
262 positioned at 25% and 75% of the scaled knee joint width in accordance with Barrios & Willson
263 ⁴¹.

264

265 **@@@FIGURE 5 NEAR HERE@@@**

266

267 The model was firstly scaled within OpenSim to account for the anthropometrics of each
268 participant, using data from the anatomical landmarks collected during the static calibration

269 trials. In accordance with Kar & Quesada,³⁵, muscle and ligament dimensions were scaled in
270 the same manner as body segments, from the static trial marker positions. Following this as
271 muscle forces are the main determinant of joint forces³⁴, muscle kinetics were quantified using
272 computed muscle control (CMC) procedure to estimate a set of muscle force patterns allowing
273 the model to replicate the required kinematics.

274

275 Then, three-dimensional ankle, medial tibiofemoral, lateral tibiofemoral and hip joint forces as
276 well as compressive patellofemoral joint forces were calculated via the joint reaction analyses
277 function within OpenSim, using the muscle forces generated from the CMC process as inputs.
278 The joint reaction analysis function in OpenSim calculates the joint loads transferred between
279 two contacting bodies, about the joint location identified during the static trial. Furthermore,
280 the three-dimensional forces calculated at the lateral and medial aspects of the tibiofemoral
281 joint via the joint reaction analysis were added together in order to also determine the total
282 tibiofemoral joint force in all three planes. In the current investigation, joint forces were
283 normalized by dividing by each participants body weight (BW).

284

285 From the above processing, peak three-dimensional ankle, lateral tibiofemoral, medial
286 tibiofemoral, total tibiofemoral and hip joint forces, and peak compressive patellofemoral
287 forces during the stance phase were extracted for statistical analyses. In addition, instantaneous
288 load rates (BW/s) for each of the aforementioned joint loads were extracted by obtaining the
289 peak increase in force between adjacent data points and joint force impulses (BW·ms) during
290 the stance phase were also calculated using a trapezoidal function.

291

292 In addition to the above, from the CMC process firstly the peak ACL force during the stance
293 phase was extracted and normalized by dividing the net values by bodyweight (BW).

294 Furthermore, the peak forces (BW) during the stance phase for the major muscles crossing the
295 knee joint were quantified and also the muscle force impulses (BW·ms) during the stance phase
296 were also extracted using a trapezoidal function. In addition, the biceps femoris long head,
297 biceps femoris short head, semitendinosus, semimembranosus muscle forces calculated via the
298 CMC process were added together to create the total hamstring muscle force. In addition, the
299 rectus femoris, vastus lateralis, vastus medialis and vastus intermedius forces calculated via the
300 CMC process were also summed to create the total quadriceps muscle force. The maximum
301 total hamstring and total quadriceps forces as well as their impulses during the stance phase
302 were extracted for statistical analysis.

303

304 In addition, the maximum ACL strain (%) was calculated by dividing the maximum ligament
305 bundle length during the dynamic trials by the resting length, which was obtained during the
306 static calibration trials³⁵ and ACL strain rate (%/s) was by obtaining the peak increase in ACL
307 strain between adjacent data points.

308

309 *Statistical analyses*

310 For each parameter/ condition, means and standard deviations were calculated and differences
311 between knee sleeve and no-sleeve conditions examined using Bayesian paired t-tests with
312 default prior scales using SPSS 27.0 software (SPSS, IBM). Bayesian factors (BF) were used
313 to explore the extent to which the data supported the alternative (H_1) hypothesis and Bayes
314 factors throughout were interpreted in accordance with the recommendations of Jeffreys⁴² with
315 values ≥ 3 indicating sufficient evidence in support of H_1 . In the interests of conciseness and
316 clarity only variables that presented with Bayes factors ≥ 3 are presented in the results section.
317 Finally, using the data collected from the subjective feedback based on participants' ratings of
318 both stability and comfort were examined using Chi-Square tests.

319

320 Experiment 2

321 *Participants*

322 Ten male (age 21.1 ± 1.13 years, height 1.77 ± 0.1 m, body mass 71.9 ± 8.6 kg and BMI =
323 22.9 ± 3.2 kg/m²) university first team level footballers volunteered to take part in the current
324 investigation.

325

326 *Procedure*

327 Kinematic information was obtained using the procedure and biomechanical modelling
328 approach outlined in experiment 1 and participants once again wore the same footwear. For
329 this experiment participants performed single leg drop jump landings with their right
330 (dominant) limb after stepping off from a 30 cm plyometric box onto the force platform in
331 order to simulate deceleration phase of landing⁴³. The landing phase of was considered to have
332 begun at foot contact (defined as > 20 N of vertical force applied to the force platform) and
333 ended at the instance of maximum knee flexion.

334

335 *Processing*

336 The same processing techniques and variables as experiment 1 were adopted.

337

338 *Statistical analyses*

339 To examine biomechanical differences between conditions and subjective preferences/ ratings
340 the same statistical analyses as experiment 1 were adopted, with the same statistical principles
341 and reporting adhered to.

342

343 **Results**

344 Experiment 1

345 @@@ TABLE 1 NEAR HERE @@@

346 @@@ TABLE 2 NEAR HERE @@@

347 @@@ TABLE 3 NEAR HERE @@@

348

349 *Ligament biomechanics*

350 For the peak ACL strain, values were larger in the knee sleeve (BF = 4.45) condition compared
351 to no-sleeve (Table 1). For the peak ACL force, values were larger in the knee sleeve (BF =
352 25.53) condition compared to no-sleeve (Table 2).

353

354 *Joint loading*

355 For the hip shear force impulse values were larger in the knee sleeve (BF = 33.31) compared
356 to no-sleeve (Table 1). Furthermore, for the hip medial force impulse values were larger in the
357 knee sleeve (BF = 7.70) compared to no-sleeve (Table 1).

358

359 For the peak lateral tibiofemoral compressive force, values were larger in the knee sleeve (BF
360 = 28.55) conditions compared to no-sleeve (Table 1). For the peak total compressive
361 tibiofemoral force, values were greater in the knee sleeve (BF = 4.04) conditions compared to
362 no-sleeve (Table 1).

363

364

365 *Joint kinematics and proprioception*

366 No differences in joint kinematics or proprioception (BF <3.0) were observed (Table 2).

367

368 *Muscle forces*

369 For peak vastus medialis force, values were larger in the knee sleeve compared to no-sleeve
370 (BF = 3.11) (Table 3). For peak gracilis force, values were larger in the no-sleeve condition
371 compared to the knee sleeve (BF = 5.56) (Table 3). Similarly, for the gracilis force integral,
372 values were larger in the no-sleeve condition compared to the knee sleeve (BF = 11.81) (Table
373 3).

374

375 Subjective ratings

376 For the subjective ratings, participants indicated that the sleeve significantly improved
377 subjective comfort ($X^2_{(2)} = 13.50, p < 0.05$) and subjective stability ($X^2_{(2)} = 8.33, p < 0.05$).

378

379 Experiment 2

380

@@@ TABLE 4 NEAR HERE @@@

381

@@@ TABLE 5 NEAR HERE @@@

382

@@@ TABLE 6 NEAR HERE @@@

383

384 Ligament biomechanics

385 No differences in ligament biomechanics (BF < 3.0) were observed (Table 4).

386

387 Joint loading

388 No differences in joint loading (BF < 3.0) were observed (Table 4).

389

390 Joint kinematics and proprioception

391 No differences in joint kinematics or proprioception (BF < 3.0) were observed (Table 5).

392

393 Muscle forces

394 No differences in muscle forces (BF < 3.0) were observed (Table 6).

395

396 Subjective ratings

397 For the ratings of comfort, participants indicated that the sleeve did not significantly influence
398 subjective comfort ($X^2_{(2)} = 1.75, p > 0.05$) or stability ($X^2_{(2)} = 3.25, p > 0.05$).

399

400 **Discussion**

401 The current investigation using a two-experiment approach, represents the first study to explore
402 the effects of prophylactic knee bracing on ACL loading/ strains alongside lower extremity
403 joint loading using musculoskeletal simulation in male and female football players. The
404 debilitating nature of ACL injuries, the high rate of re-injury and the incidence of degenerative
405 joint disease secondary to ACL injury, means that this study may provide important
406 information necessary to inform future prevention strategies and insight into the cumulative
407 biomechanical effects of prophylactic knee braces.

408

409 In relation to the ACL, experiment 1 showed that ACL loading and ACL strain were larger in
410 the knee sleeve compared to no-sleeve. This observation opposes those of Sinclair et al.,³¹ and
411 Sinclair et al.,³² who showed that prophylactic knee bracing attenuated knee joint soft tissue
412 loading at the patellar tendon and ACL itself. Mechanically, aetiological analyses have shown
413 that ACL injuries occur when the ligament itself experiences excessive tensile forces and
414 strains¹². Given the increases in these parameters shown in experiment 1, it appears that
415 prophylactic knee bracing akin to that examined in this study may increase the risk from the
416 ligamentous parameters linked to the aetiology of injury. Therefore, during the sports specific
417 movements examined in experiments 1 and 2, the findings do not support the utilization of
418 prophylactic knee bracing for the attenuation ACL injuries.

419

420 At the tibiofemoral joint, experiment 1 indicated that lateral and total tibiofemoral compressive
421 loading was larger in the knee sleeve. As no-differences in medial tibiofemoral compartment
422 loading were found it can be concluded that differences in total tibiofemoral loading were
423 mediated through increases at the lateral tibiofemoral compartment. Whilst prophylactic knee
424 bracing has been shown to attenuate tibiofemoral loading quantified using the peak knee
425 adduction moment during a badminton lung³⁰, there has yet to be an examination of the effects
426 of knee bracing on lateral tibiofemoral kinetics. Nonetheless, despite medial tibiofemoral
427 disorders being far more commonplace⁴⁴, the aetiology of joint degenerative pathologies is
428 linked to excessive and habitual mechanical loading⁴⁵. As such, experiment 1 indicates that
429 the knee sleeve may increase the risk from the biomechanical mechanisms linked to the
430 initiation of lateral tibiofemoral degeneration during the cut movement. Therefore, similar to
431 the conclusions in relation to the ACL, the findings do not support the utilization of
432 prophylactic knee bracing for the attenuation of knee joint injuries in male and female
433 footballers during 45°cut and single leg landing conditions.

434

435 At the hip joint, the findings from experiment 1 showed that both the shear and medial force
436 impulses were significantly larger in the knee sleeve condition compared to no-sleeve. This
437 observation supports the principles of the walking study shown by Toriyama et al.,⁴⁶, in that a
438 knee brace significantly attenuated hip joint kinetics of the ipsilateral side. This investigation
439 therefore highlights that knee sleeves affect joint mechanics in addition to those experienced
440 by the knee joint itself. Thus, it is recommended that future analyses concerning knee braces,
441 examine more than knee joint biomechanics in order to obtain a more cumulative representation
442 of their potential prophylactic effects. Regardless, as the aetiology of hip joint degeneration is

443 linked to the magnitude and frequency at which the applied mechanical loads are experienced
444 ⁴⁵, experiment 1 indicates that the knee sleeve may enhance the risk from the kinetic
445 mechanisms linked to the initiation of hip joint degeneration.

446

447 Previous systematic analyses have proposed that prophylactic knee braces promote and
448 facilitate safer landing biomechanics during functional athletic tasks by promoting an increased
449 sensation of knee joint stability ⁴⁷. However, the subjective and proprioceptive ratings from
450 both experiments in the current investigation provide only partial support for this notion.
451 Experiment 1 showed that the knee sleeve enhanced subjective knee joint stability yet in
452 experiment 2 there were no perceptual alterations as a function of the sleeve, and neither
453 investigation showed any improvement in knee joint proprioception. It is proposed that knee
454 braces enhance knee joint stability and proprioception by stimulating sense receptors in the
455 skin mediated through compression provided by the brace itself ⁴⁷. However, the findings from
456 experiment 1 do not appear to support this, as whilst improvements in perceived stability were
457 shown, this did not translate into positive changes in knee biomechanics. It has been speculated
458 previously that prophylactic sleeves do not provide sufficient compression to alter knee
459 stability and proprioception sufficiently to mediate alterations in dynamic knee biomechanics
460 ²⁹. Therefore, although compression provided via the knee sleeve was not examined as part of
461 the current investigation, an interesting avenue for future analyses may be to explore devices
462 that provide different levels of compression in regards to their prophylactic efficacy.

463

464 A potential limitation to both experiments undertaken as part of the current investigation is the
465 mechanism by which the musculoskeletal simulation-based analyses were completed. The
466 CMC process, although an effective and robust tool for the quantification of muscle and soft

467 tissue kinetics utilized in previous analyses to simulate ACL mechanics ³⁵, can be limited in its
468 ability to quantify specific muscle coordination during dynamic tasks ⁴⁸. Furthermore, that the
469 ACL was not modelled with sex specificity in regard to its anatomy and scaling may serve as
470 a drawback to this investigation. Although such an approach has yet to be developed within the
471 simulation based musculoskeletal modelling literature; as the ACL contributes pointedly to
472 knee mechanics, incorporation of sex-specific ligament modelling may improve the efficacy of
473 musculoskeletal simulation analyses. Finally, that only relatively modest sample sizes were
474 utilized in both experiments may have limited statistical power and alternate statistical
475 observations may have arisen as a function of enhanced Bayes factors with the inclusion of
476 additional participants ⁴⁹.

477

478 **Conclusion**

479 The current investigation adds to the literature by exploring via a two-experiment investigation,
480 the effects of prophylactic knee bracing on ACL loading/ strains and lower extremity joint
481 biomechanics using a musculoskeletal simulation-based approach in male and female
482 footballers. This study importantly showed in experiment 1 that ACL loading/ strain, lateral
483 and total tibiofemoral compressive forces as well as hip joint shear and medial forces were
484 greater in the knee sleeve condition and in experiment 2 that there were no statistical effects of
485 the knee sleeve. Therefore, the findings from the current investigation suggest that the
486 prophylactic knee sleeve examined in the current investigation does not appear to reduce the
487 biomechanical parameters linked to the aetiology of knee pathologies in male/ female
488 footballers.

489

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619

620 **Tables**

621 Table 1: ACL and joint forces (Means ± standard deviations) for each knee sleeve condition – from
 622 experiment 1.

	Knee sleeve		No-sleeve	
	Mean	SD	Mean	SD
Peak ACL force (BW)	1.19	0.36	0.94	0.33
Peak ACL strain (%)	13.57	4.84	10.26	2.38
Peak ACL strain (%/s)	75.37	10.96	80.87	12.39
Peak hip compressive force (BW)	9.97	1.84	9.80	1.74
Hip compressive impulse (BW·ms)	1652.58	433.36	1708.26	452.87
Peak hip shear force (BW)	2.74	1.35	2.49	1.21
Hip shear impulse (BW·ms)	194.20	306.42	92.70	286.98
Hip peak medio-lateral force (BW)	4.93	1.15	5.85	1.10
Hip medio-lateral impulse (BW·ms)	702.52	301.03	830.84	310.57
Peak patellofemoral compressive force (BW)	10.08	2.45	10.17	3.00
Patellofemoral compressive impulse (BW·ms)	1350.34	465.84	1414.64	531.04
Peak medial tibiofemoral condyle compressive force (BW)	7.22	1.50	7.11	1.51
Medial tibiofemoral condyle compressive impulse (BW·ms)	1052.14	284.94	1021.79	301.41
Peak medial tibiofemoral condyle shear force (BW)	3.84	1.03	4.30	0.76
Medial tibiofemoral condyle shear impulse (BW·ms)	532.59	153.16	641.38	149.64
Peak medial tibiofemoral medio-lateral force (BW)	2.22	1.41	1.96	1.03
Peak medial tibiofemoral medio-lateral impulse (BW·ms)	306.96	196.94	265.22	195.17
Peak lateral tibiofemoral condyle compressive force (BW)	4.70	0.95	4.20	1.14
Lateral tibiofemoral condyle compressive impulse (BW·ms)	698.05	273.28	660.11	285.97
Peak lateral tibiofemoral condyle shear force (BW)	2.30	0.71	2.41	0.90
Lateral tibiofemoral condyle shear impulse (BW·ms)	316.40	149.65	334.28	163.53
Peak lateral tibiofemoral medio-lateral force (BW)	1.88	0.83	1.68	0.50
Peak lateral tibiofemoral medio-lateral impulse (BW·ms)	265.43	134.46	231.05	94.07
Peak total tibiofemoral compressive force (BW)	11.73	2.34	11.08	2.49

Total tibiofemoral compressive impulse (BW·ms)	1750.18	534.36	1681.90	569.46
Peak total tibiofemoral shear force (BW)	5.87	1.32	6.45	1.15
Total tibiofemoral shear impulse (BW·ms)	849.00	209.43	975.66	268.93
Peak total tibiofemoral medio-lateral force (BW)	3.79	2.27	3.31	1.48
Peak total tibiofemoral medio-lateral impulse (BW·ms)	572.39	318.22	496.27	268.97
Peak ankle compressive force (BW)	10.36	1.48	10.08	2.13
Ankle compressive impulse (BW·ms)	1525.02	387.94	1453.99	408.65
Peak ankle shear force (BW)	3.14	0.91	3.20	1.24
Ankle shear impulse (BW·ms)	191.72	237.15	100.94	255.53
Peak ankle medio-lateral force (BW)	3.96	3.96	3.94	3.94
Ankle medio-lateral impulse (BW·ms)	550.48	245.11	510.37	188.78

623 Notes: bold text = statistical difference between knee-sleeve and no-sleeve conditions (BF >3.00).

624

625 Table 2: Knee joint kinematics (Means ± standard deviations) for each knee brace condition – from
626 experiment 1.

	Knee sleeve		No-sleeve	
	Mean	SD	Mean	SD
Peak knee flexion (°)	60.94	11.63	60.08	9.52
Peak knee abduction (°)	11.44	5.99	13.33	8.81
Peak knee internal rotation (°)	10.04	6.48	6.06	7.86
Peak knee flexion velocity (°/s)	505.39	70.22	464.80	113.63
Peak knee abduction velocity (°/s)	205.60	127.17	161.93	69.48
Peak knee internal rotation velocity (°/s)	288.25	150.05	308.87	108.13
Proprioception angular error (°)	3.93	1.93	4.23	1.88

627

628 Table 3: Muscle forces (Means ± standard deviations) for each knee sleeve condition – from
629 experiment 1.

	Knee sleeve		No-sleeve	
	Mean	SD	Mean	SD
Peak biceps femoris long head force (BW)	0.49	0.31	0.46	0.33
Biceps femoris long head impulse (BW·ms)	39.60	48.66	31.17	31.03
Peak biceps femoris short-head force (BW)	0.79	0.29	0.83	0.26
Biceps femoris short head impulse (BW·ms)	60.11	36.24	59.85	28.25
Peak gracilis force (BW)	0.14	0.06	0.21	0.10
Gracilis impulse (BW·ms)	7.61	5.15	10.27	5.47
Peak lateral gastrocnemius force (BW)	1.11	0.25	1.03	0.36
Lateral gastrocnemius impulse (BW·ms)	81.85	29.55	75.65	34.28
Peak medial gastrocnemius force (BW)	2.18	0.62	2.41	0.57
Medial gastrocnemius impulse (BW·ms)	166.00	65.81	172.83	57.86
Peak rectus femoris force (BW)	2.83	0.65	2.87	0.57
Rectus femoris impulse (BW·ms)	358.71	165.51	381.55	178.20

Peak semimembranosus force (BW)	0.84	0.46	0.80	0.41
Semimembranosus impulse (BW·ms)	59.06	33.27	55.53	31.33
Peak semitendinosus force (BW)	0.27	0.10	0.27	0.11
Semitendinosus impulse (BW·ms)	15.34	7.51	15.06	7.36
Peak total hamstring force (BW)	1.80	0.73	1.61	0.61
Total hamstring impulse (BW·ms)	174.11	89.74	161.61	75.78
Peak total quadriceps force (BW)	9.80	1.92	9.39	2.21
Total quadriceps impulse (BW·ms)	1412.64	397.21	1417.13	437.73
Peak vastus intermedius force (BW)	2.61	0.48	2.46	0.70
Vastus intermedius impulse (BW·ms)	309.22	75.38	304.77	95.09
Peak vastus lateralis force (BW)	3.97	0.68	3.77	0.97
Vastus lateralis impulse (BW·ms)	457.30	121.75	450.42	149.08
Peak vastus medialis force (BW)	2.43	0.49	2.27	0.68
Peak vastus medialis impulse (BW·ms)	287.41	72.86	280.39	88.58

630 Notes: bold text = statistical difference between knee-sleeve and no-sleeve conditions (BF >3.00).

631

632 Table 4: ACL and joint forces (Means ± standard deviations) for each knee sleeve condition – from
633 experiment 2.

	Knee sleeve		No-sleeve	
	Mean	SD	Mean	SD
Peak ACL force (BW)	0.97	0.18	0.92	0.11
Peak ACL strain (%)	12.83	3.06	11.71	1.35
Peak ACL strain (%/s)	105.94	11.57	106.67	19.57
Peak hip compressive force (BW)	9.82	2.00	10.16	1.55
Hip compressive impulse (BW·ms)	1450.31	295.85	1521.78	273.71
Peak hip shear force (BW)	2.19	0.48	2.52	0.69
Hip shear impulse (BW·ms)	302.29	118.70	368.95	153.17
Hip peak medio-lateral force (BW)	1.39	0.66	1.50	0.75
Hip medio-lateral impulse (BW·ms)	178.76	109.51	194.07	79.30
Peak patellofemoral compressive force (BW)	8.13	1.24	8.01	1.98
Patellofemoral compressive impulse (BW·ms)	1309.86	428.92	1337.48	568.47
Peak medial tibiofemoral condyle compressive force (BW)	6.83	1.61	6.80	1.04
Medial tibiofemoral condyle compressive impulse (BW·ms)	1042.57	221.16	1096.50	355.52
Peak medial tibiofemoral condyle shear force (BW)	2.69	0.26	2.70	0.52
Medial tibiofemoral condyle shear impulse (BW·ms)	424.84	131.27	409.98	140.09
Peak medial tibiofemoral medio-lateral force (BW)	0.92	0.30	0.82	0.27
Peak medial tibiofemoral medio-lateral impulse (BW·ms)	136.67	51.50	132.54	72.57
Peak lateral tibiofemoral condyle compressive force (BW)	5.22	0.95	4.65	0.56
Lateral tibiofemoral condyle compressive impulse (BW·ms)	618.42	122.87	639.73	153.60
Peak lateral tibiofemoral condyle shear force (BW)	1.84	0.38	1.82	0.46
Lateral tibiofemoral condyle shear impulse (BW·ms)	274.89	96.87	270.80	118.40
Peak lateral tibiofemoral medio-lateral force (BW)	0.32	0.15	0.30	0.08
Peak lateral tibiofemoral medio-lateral impulse (BW·ms)	27.63	19.11	25.58	17.72

Peak total tibiofemoral compressive force (BW)	11.27	1.97	10.63	0.97
Total tibiofemoral compressive impulse (BW·ms)	1660.99	312.75	1736.22	496.97
Peak total tibiofemoral shear force (BW)	4.42	0.60	4.37	0.92
Total tibiofemoral shear impulse (BW·ms)	699.73	222.50	680.78	255.84
Peak total tibiofemoral medio-lateral force (BW)	1.22	0.43	1.06	0.33
Peak total tibiofemoral medio-lateral impulse (BW·ms)	164.30	65.60	158.13	83.70
Peak ankle compressive force (BW)	8.69	1.29	8.97	1.48
Ankle compressive impulse (BW·ms)	1393.27	219.68	1442.64	333.20
Peak ankle shear force (BW)	2.33	0.57	1.99	1.29
Ankle shear impulse (BW·ms)	270.75	164.04	226.61	209.43
Peak ankle medio-lateral force (BW)	0.68	0.34	0.77	0.63
Ankle medio-lateral impulse (BW·ms)	62.85	53.41	67.67	54.34

634

635 Table 5: Knee joint kinematics (Means \pm standard deviations) for each knee brace condition – from
636 experiment 2.

	Knee sleeve		No-sleeve	
	Mean	SD	Mean	SD
Peak knee flexion (°)	65.71	7.89	66.80	8.46
Peak knee abduction (°)	4.93	3.62	3.54	3.95
Peak knee internal rotation (°)	1.66	8.46	1.78	4.35
Peak knee flexion velocity (°/s)	639.08	17.85	641.84	52.57
Peak knee abduction velocity (°/s)	102.89	41.47	159.01	50.95
Peak knee external rotation velocity (°/s)	206.35	102.86	180.05	71.63
Proprioception angular error (°)	4.13	2.39	4.42	2.15

637

638 Table 6: Muscle forces (Means \pm standard deviations) for each knee sleeve condition – from
639 experiment 2.

	Knee sleeve		No-sleeve	
	Mean	SD	Mean	SD
Peak biceps femoris long head force (BW)	0.37	0.16	0.53	0.21
Biceps femoris long head impulse (BW·ms)	39.07	33.30	44.42	19.98
Peak biceps femoris short-head force (BW)	0.37	0.19	0.55	0.27
Biceps femoris short head impulse (BW·ms)	19.88	8.22	33.72	24.87
Peak gracilis force (BW)	0.06	0.03	0.06	0.03
Gracilis impulse (BW·ms)	3.22	1.38	3.62	2.04
Peak lateral gastrocnemius force (BW)	0.50	0.16	0.73	0.31
Lateral gastrocnemius impulse (BW·ms)	44.59	16.82	62.47	32.80
Peak medial gastrocnemius force (BW)	1.20	0.34	1.69	0.66
Medial gastrocnemius impulse (BW·ms)	93.97	55.19	114.90	46.88
Peak rectus femoris force (BW)	1.96	0.33	1.90	0.36
Rectus femoris impulse (BW·ms)	161.77	40.99	176.52	36.50
Peak semimembranosus force (BW)	0.45	0.19	0.71	0.36
Semimembranosus impulse (BW·ms)	35.81	23.82	50.87	30.94

Peak semitendinosus force (BW)	0.18	0.07	0.18	0.06
Semitendinosus impulse (BW·ms)	10.04	4.61	13.89	6.55
Peak total hamstring force (BW)	1.21	0.41	1.68	0.48
Total hamstring impulse (BW·ms)	104.80	64.93	142.90	52.45
Peak total quadriceps force (BW)	7.95	1.25	7.42	1.40
Total quadriceps impulse (BW·ms)	1284.33	360.81	1285.00	532.72
Peak vastus intermedius force (BW)	2.04	0.36	1.85	0.49
Vastus intermedius impulse (BW·ms)	319.08	100.44	315.40	144.75
Peak vastus lateralis force (BW)	3.15	0.37	2.96	0.76
Vastus lateralis impulse (BW·ms)	513.36	159.53	502.71	227.52
Peak vastus medialis force (BW)	1.85	0.35	1.76	0.46
Peak vastus medialis impulse (BW·ms)	290.12	95.97	290.38	135.55

640

641 **Figure labels**

642 **Figure 1: Experimental knee sleeve.**

643 **Figure 2: Experimental laboratory set-up with motion capture system cameras numbered**
644 **according to the laboratory system and force platform (FP). Approach (A) and cut (C)**
645 **directions are labelled with arrows showing participants direction of travel as part of the 45°**
646 **cut movement.**

647 **Figure 3: a. Experimental marker locations and b. trunk, pelvis, thigh, shank and foot segments,**
648 **with segment co-ordinate system axes (R = right & L = left), (TR = trunk, P = pelvis, T = thigh,**
649 **S = shank & F = foot), (X = sagittal, Y = coronal & Z = transverse planes).**

650 **Figure 4: Weight-bearing knee joint position sense test from a. frontal and b. sagittal**
651 **viewpoints.**

652 **Figure 5: a. Experimental Opensim model in full and b. with only the ACL bundles visible.**