Sport Sciences for Health

Effects of second generation and indoor sports surfaces on knee joint kinetics and kinematics during 45° and 180° cutting manoeuvres; and exploration using statistical parametric mapping and Bayesian analyses.
--Manuscript Draft--

Manuscript Number:	SSFH-D-19-00207R2
Full Title:	Effects of second generation and indoor sports surfaces on knee joint kinetics and kinematics during 45° and 180° cutting manoeuvres; and exploration using statistical parametric mapping and Bayesian analyses.
Article Type:	Original Article
Funding Information:	
Abstract:	Purpose: The aim of the current investigation was to examine the influence of second generation (2G) and indoor surfaces on knee joint kinetics, kinematics, frictional and muscle force parameters during 45° and 180° change of direction movements using statistical parametric mapping (SPM) and Bayesian analyses. Methods: Twenty male participants performed 45° and 180° change of direction movements on 2G and indoor surfaces. Lower limb kinematics were collected using an eight-camera motion capture system and ground reaction forces were quantified using an embedded force platform. ACL, patellar tendon and patellofemoral loading was examined via a musculoskeletal modelling approaches and the frictional properties of the surfaces were examined using ground reaction force information. Differences between surfaces were examined using SPM and Bayesian analyses. Results: Both SPM and Bayesian analyses showed that ACL loading parameters were greater in the 2G condition in relation to the indoor surface. Conversely, SPM and Bayesian analyses confirmed that patellofemoral/ patellar tendon loading alongside the coefficient of friction and peak rotational moment were larger in the indoor condition compared to the 2G surface. Conclusions: This study indicates that the indoor surface may improve change of direction performance owing to enhanced friction at the shoe-surface interface but augment the risk from patellar tendon/ patellofemoral injuries; whereas the 2G condition may enhance the risk from ACL pathologies.
Corresponding Author:	Jonathan Sinclair, PhD, Senior Fellow HEA University of Central Lancashire UNITED KINGDOM
Corresponding Author Secondary Information:	
Corresponding Author's Institution:	University of Central Lancashire
Corresponding Author's Secondary Institution:	
First Author:	Jonathan Sinclair, PhD, Senior Fellow HEA
First Author Secondary Information:	
Order of Authors:	Jonathan Sinclair, PhD, Senior Fellow HEA
	Naomi Liles, BSc (Hons)
	Paul John Taylor, PhD
	Thomas Glenn, BSc (Hons), MSc
Order of Authors Secondary Information:	
Author Comments:	
Response to Reviewers:	Please see attached file.

1 **Effects of second generation and indoor sports surfaces on knee joint kinetics and**

2 **kinematics during 45° and 180° cutting manoeuvres; and exploration using statistical**

3 **parametric mapping and Bayesian analyses.**

4 *Jonathan Sinclair[1], Naomi Liles[1], Paul John Taylor[2] & Thomas Glenn[1]*

5 1. *Centre for Applied Sport and Exercise Sciences, School of Sport and Health Sciences,*

6 *Faculty of Health and Wellbeing, University of Central Lancashire.*

7 2. *School of Psychology, Faculty of Science & Technology, University of Central*

8 *Lancashire.*

9 **Correspondence Address:**

10 Dr. Jonathan Sinclair,

11 Centre for Applied Sport Exercise and Nutritional Sciences

12 Faculty of Health & Wellbeing

13 University of Central Lancashire,

14 Preston

15 Lancashire

16 PR1 2HE.

17 e-mail: jksinclair@uclan.ac.uk

18 **Keywords:** Biomechanics; surface; statistical parametric mapping; Bayesian; sport.

19

20 **Abstract**

21 **Purpose:** The aim of the current investigation was to examine the influence of second

22 generation (2G) and indoor surfaces on knee joint kinetics, kinematics, frictional and muscle

23 force parameters during 45° and 180° change of direction movements using statistical

24 parametric mapping (SPM) and Bayesian analyses.

25   **Methods:** Twenty male participants performed 45° and 180° change of direction movements

26   on 2G and indoor surfaces. Lower limb kinematics were collected using an eight-camera

27   motion capture system and ground reaction forces were quantified using an embedded force

28   platform. ACL, patellar tendon and patellofemoral loading was examined via a musculoskeletal

29   modelling approaches and the frictional properties of the surfaces were examined using ground

30   reaction force information. Differences between surfaces were examined using SPM and

31   Bayesian analyses.

32   **Results:** Both SPM and Bayesian analyses showed that ACL loading parameters were greater

33   in the 2G condition in relation to the indoor surface. Conversely, SPM and Bayesian analyses

34   confirmed that patellofemoral/ patellar tendon loading alongside the coefficient of friction and

35   peak rotational moment were larger in the indoor condition compared to the 2G surface.

36   **Conclusions:** This study indicates that the indoor surface may improve change of direction

37   performance owing to enhanced friction at the shoe-surface interface but augment the risk from

38   patellar tendon/ patellofemoral injuries; whereas the 2G condition may enhance the risk from

39   ACL pathologies.

40

41   **Introduction**

42   The benefits of physical activity/ sport are unequivocal [1] and initiatives to improve

43   participation are commonplace [2]. However, sports/ physical activity is associated with a high

44   incidence of injuries [3, 4]. The annual cost of treating sports injuries in high school athletes

45   alone is estimated to be >$2 billion [4], with 1/5 school children absent at least one day per year

46   [6] and 1/3 working adults losing at least one day a year to sports injuries [6].

47

48   Importantly, Hootman et al., [7] showed in their aetiological examination of 15 different sports,

49   that the lower extremities were the most common location for injury. Specifically, the knee

50   joint is the most commonly injured musculoskeletal structure in athletes, accounting for over

51   30% of all reported sports injuries [8]. The most frequently reported knee condition in sports

52   medicine clinics is patellofemoral pain which has a prevalence cited between 22.7-28.9% [9],

53   and manifests as dull retropatellar pain, aggravated by activities that frequently and excessively

54   load the joint [10]. Chronic patellar tendinopathy is also a common musculoskeletal condition

55   that presents clinically as localised pain at the proximal tendon attachment [11]. Aetiological

56   analyses have shown that the incidence of patellar tendinopathy may be as high as 36%, with

57   this specific condition accounting for as many as 25% of all soft tissue injuries [12].

58   Tendinopathy is mediated through excessively forces at the patellar tendon itself, with failed

59   reparative response due to insufficient rest between loading exposures [11]. Similarly, the

60   anterior cruciate ligament (ACL) is the most frequently reported acute sports injury [13], with

61   over 175,000 ACL reconstruction surgeries being performed each year in the US alone [14].

62   ACL injuries are predominantly non-contact in nature, whereby the ligament becomes

63   compromised in the absence of physical contact between athletes [15]. Mechanically, ACL

64   injuries occur when the ligament experiences excessive tensile forces [16].

65

66   Given the prevalence and debilitating nature of sports injuries, considerable research attention

67   has been devoted to modifying the underlying mechanisms linked to the aetiology of common

68   sports-related pathologies. It has been strongly advocated that the properties of sports playing

69   surfaces can influence both the performance of athletes and the likelihood of injury occurrence

70   [17]. Traditionally most sports were played on natural surfaces, however, owing to climatic and

71   economic factors, artificial alternatives have become increasing popular over the past 30 years,

72   with synthetic grass and indoor surfaces being the most commonly encountered [18]. Many

73   athletic disciplines involve sprinting, stopping and rapid changes in movement direction [19].

74   Frictional torque generated at the shoe-surface interface means that the knee may be subjected

75 to excessive stresses when rapid directional changes are undertaken [20]. Therefore, the level

76 of traction between the shoe and surface is one of the most commonly cited factors influencing

77 lower limb injury occurrence [21].

78

79 There are concerns that some of the mechanical properties of artificial sports-surfaces may be

80 associated with acute and chronic knee injuries in relation to natural surface, and there is

81 evidence from descriptive epidemiological analyses to support this notion [22]. However,

82 clinical, biomechanical and epidemiological analyses have focused heavily on the differences

83 between performing on natural vs. synthetic surfaces, and there has yet to be a published

84 investigation examining different synthetic surfaces on the biomechanical mechanisms linked

85 specifically to the aetiology of knee pathologies.

86

87 Furthermore, whilst aetiological literature has through prospective and retrospective analyses

88 been able to identify the risk factors linked to the aetiology of knee pathologies, these

89 biomechanical parameters are explored in the scientific literature through discrete point analysis

90 [23]. For time normalized biomechanical parameters, statistical parametric mapping (SPM)

91 may represent a more efficacious process, as it is able to examine an entire time-based data

92 sequence and reduces the likelihood of a type II error by eliminating the need for multiple

93 analyses [23]. Similarly, in recent years Bayesian assessments have gained considerable

94 acceptance and practicability, although in spite of their prospective benefits [24], their

95 utilization in biomechanical analyses remains limited. To date there has yet to be a

96 biomechanical investigation examining the effects of different synthetic surfaces on the

97 biomechanical parameters linked to the aetiology of knee pathologies using a concurrent SPM

98 and Bayesian approach.

99

100 Therefore, the aim of the current investigation was to examine the influence of second

101 generation (2G) and indoor surfaces on knee joint kinetics, kinematics, frictional and muscle

102 force parameters during 45° and 180° change of direction movements using SPM and Bayesian

103 analyses.

104

105 **Methods**

106 *Participants*

107 Twenty male recreational athletes (age = 23.00±2.51years, stature = 176.22±8.36cm and mass

108 = 76.79±10.60kg) volunteered to take part in this study. The procedure utilized for this

109 investigation was approved by an institutional ethical committee. All participants were free

110 from musculoskeletal pathology at the time of data collection and had not previously undergone

111 knee surgery. Written informed consent was obtained in accordance with the principles outlined

112 in the Declaration of Helsinki.

113

114 *Surfaces*

115 The data collection took place in an indoor biomechanics laboratory. The indoor surface

116 (MondoSport Ramflex, Mondo, Italy) had a total thickness of 6 mm, with a vulcanized rubber

117 construction. The indoor surface was comprised of a 2 mm surface layer and a 4 mm base layer

118 and was mounted over an underlying concrete surface. The 2G surface utilized for this

119 investigation was an 8 mm polyethylene, synthetic turf. For the 2G surface condition, the turf

120 was strong affixed to the existing laboratory surface and force platform using double-sided

121 carpet tape. Following the completion of their data collection protocol, participants were asked

122 to subjectively indicate which surface that they preferred, and which surface they felt provided

123 more traction.

124

Procedure

126   Participants completed five repeats of two sport-specific movements 45° and 180° change of

127   direction in the two surface conditions. To control for any order effects, the order in which

128   participants performed in each movement and surface condition was counterbalanced.

129   Kinematic information was obtained using an eight-camera motion capture system (Qualisys

130   Medical AB, Goteburg, Sweden) using a capture frequency of 250 Hz. Dynamic calibration of

131   the system was performed before each data collection session. Calibrations producing residuals

132   <0.85 mm and points above 4000 in all cameras were considered acceptable. To measure kinetic

133   information an embedded piezoelectric force platform (Kistler National Instruments, Model

134   9281CA) operating at 1000 Hz was utilised. The kinetic and kinematic information were

135   synchronously obtained using an analogue to digital board and interfaced using Qualisys track

136   manager.

137

138   Lower extremity segments were modelled in 6 degrees of freedom using the calibrated

139   anatomical systems technique [25]. To define the segment co-ordinate axes of the right foot,

140   shank and thigh, retroreflective markers were placed unilaterally onto the 1st metatarsal, 5th

141   metatarsal, calcaneus, medial and lateral malleoli, medial and lateral epicondyles of the femur.

142   To define the pelvis segment further markers were positioned onto the anterior (ASIS) and

143   posterior (PSIS) superior iliac spines. Carbon fiber tracking clusters were positioned onto the

144   shank and thigh segments (Figure 1). The foot was tracked using the 1st metatarsal, 5th

145   metatarsal and calcaneus markers and the pelvis using the ASIS and PSIS markers. The centers

146   of the ankle and knee joints were delineated as the mid-point between the malleoli and femoral

147   epicondyle markers, whereas the hip joint centre was obtained using the positions of the ASIS

148   markers. Static calibration trials (not normalized to static trial posture) were obtained in each

149   footwear allowing for the anatomical markers to be referenced in relation to the tracking

150 markers/ clusters. The Z (transverse) axis was oriented vertically from the distal segment end

151 to the proximal segment end. The Y (coronal) axis was oriented in the segment from posterior

152 to anterior. Finally, the X (sagittal) axis orientation was determined using the right-hand rule

153 and was oriented from medial to lateral (Figure 2).

154

155   @@@**FIGURE 1 NEAR HERE**@@@

156   @@@**FIGURE 2 NEAR HERE**@@@

157

158 Data were collected during the 45° and 180° change of direction movements as described below:

159

160 **45° change of direction**

161 Participants completed 45° change of direction movements using an approach velocity of 4.0

162 m/s ± 5% striking the force platform with their right (dominant) limb. Cut angles were measured

163 from the centre of the force plate and the corresponding line of movement was delineated using

164 masking tape so that it was clearly evident to participants. The stance phase of this movement

165 was defined as the duration over > 20 N of vertical ground reaction force (GRF) was applied to

166 the force platform.

167

168 **180° change of direction**

169 Participants completed 180° change of direction movements using an approach velocity of 4.0

170 m/s ± 5% striking the force platform with their right (dominant) limb, then returning in the

171 initial direction of travel. The stance phase of this movement was defined as the duration over

172 > 20 N of vertical GRF was applied to the force platform.

173

174 *Processing*

175 Dynamic trials were digitized using Qualisys Track Manager in order to identify anatomical

176 and tracking markers then exported as C3D files to Visual 3D (C-Motion, Germantown, MD,

177 USA). All data were linearly normalized to 100% of the stance phase. GRF and kinematic data

178 were smoothed using cut-off frequencies of 50 and 12 Hz with a low-pass Butterworth 4th

179 order zero lag filter [26]. Three-dimensional kinematics of the knee were calculated using an

180 XYZ cardan sequence of rotations (where X=sagittal plane; Y=coronal plane and Z=transverse

181 anatomical planes). Joint moments were computed using Newton–Euler inverse-dynamics,

182 allowing net knee joint moments to be calculated. To quantify joint moments segment mass,

183 segment length, GRF and angular kinematics were utilized.

184

185 Patellofemoral loading was quantified using a model adapted from van Eijden et al., [27], in

186 accordance with the protocol of Willson et al., [28] in that co-contraction of the knee flexor

187 musculature was accounted for. This musculoskeletal model has been shown to be sufficiently

188 sensitive to resolve differences between different footwear [29], across different foot orthotic

189 configurations [30], different prophylactic knee sleeves [31], between sexes [32, 33] and

190 between those with and without patellofemoral pain [34]. Hamstring and gastrocnemius forces

191 were calculated in accordance with previously established procedures [35]. Hamstring and

192 gastrocnemius forces (N) were multiplied by their moment arms relative to the knee flexion

193 angle [36], and then summed to generate a knee flexor moment. The knee flexor moment was

194 added to the net knee extensor moment quantified using inverse dynamics and divided by the

195 quadriceps moment arm [27], to obtain quadriceps force (N) adjusted for co-contraction of the

196 knee flexors. From the above processing quadriceps and hamstring force (N·s) impulses during

197 the stance phase were extracted using a trapezoidal function. Quadriceps and hamstring force

198 (N/s) load rates were also extracted by obtaining the peak increase in force between adjacent

199 data points.

200

Patellofemoral force (N) was quantified in accordance with the protocol of van Eijden et al., [27]. Patellofemoral joint stress (MPa) was quantified by dividing the patellofemoral force by the patellofemoral contact area. Patellofemoral contact areas were obtained in accordance with the sex specific data of Besier et al., [37]. From the above processing patellofemoral force (N·s) and stress (MPa·s) impulses during the stance phase were extracted using a trapezoidal function. Patellofemoral force (N/s) and stress (MPa/s) load rates were also extracted by obtaining the peak increase in force/ stress between adjacent data points using the first derivative function in Visual 3D.

In addition, patellar tendon loading was quantified using a musculoskeletal model similarly adapted from Janssen et al., [38]. This model has been shown to be sufficiently sensitive to resolve differences in patellar tendon kinetics between different prophylactic knee sleeves [39], between sexes [40], different rehabilitation mechanisms [41] and between dominant and non-dominant limbs [42]. Once again, the derived knee flexor moment was added to the net knee extensor moment quantified using inverse dynamics, and then divided by the moment arm of the patellar tendon, generating the patellar tendon force. The sex specific tendon moment arms were quantified using the data of Herzog & Read, [43]. From the above processing, patellar tendon force (N·s) impulse during the stance phase was extracted using a trapezoidal function. Patellar tendon load rate (N/s) and was also extracted by obtaining the peak increase in force between adjacent data points using the first derivative function in Visual 3D.

ACL loading was similarly quantified using a musculoskeletal modelling approach as described and validated by Dai and Yu, [44]. This approach has been shown to be sufficiently sensitive to resolve differences in ACL force during different movements [44], different

225 prophylactic knee sleeves [45], between sexes [46] and also as a function of different athletic

226 footwear [20]. The face validity of this current model has been evaluated from two key aspects

227 in the scientific literature. Firstly, Dai and Yu, [44] showed that the model exhibited a high

228 level of consistency with values provided from in vivo ACL loading [47]. Secondly, the timing

229 of ACL rupture in dynamic tasks occurs ≤50 ms following initial foot contact [48]. The timing

230 of the peak ACL force estimated using this model by Dai and Yu, [44] and Sinclair and Taylor

231 [45] shown to be <50 ms, is therefore consistent with this data and further supports the face

232 validity of the model. From the above processing, ACL force (N·s) impulse during the stance

233 phase was extracted using a trapezoidal function. ACL load rate (N/s) and was also extracted

234 by obtaining the peak increase in force between adjacent data points. Further, to the above the

235 knee abduction moment impulse (Nm·s) during the stance phase was extracted using a

236 trapezoidal function and the abduction moment load rate (Nm/s) was also extracted by

237 obtaining the peak increase between adjacent data points using the first derivative function in

238 Visual 3D.

239

240 Finally, the loading rates (N/s) of the vertical and braking GRFs were also extracted by

241 obtaining the peak increase in vertical and anterior-posterior GRF between adjacent data points.

242 Furthermore, the peak translation coefficient of friction (μ) of each footwear was determined

243 from the ratio of horizontal and vertical force components during the initial period of shoe

244 motion [20]. The peak rotational moment of the GRF (Nm) was used to describe the rotational

245 friction characteristics of the footwear [49].

246

247 Following this, three-dimensional knee joint kinematics, vertical GRF, anterior posterior GRF,

248 quadriceps force, hamstring force, patellofemoral force, patellofemoral stress, patellar tendon,

249 ACL and knee abduction moment parameters were extracted during the entire stance phase and

250 time normalized to 101 data points using linear interpolation for each participant.

251

252 *Statistical analyses*

253 Differences across the entire stance phase were examined using 1-dimensional statistical

254 parametric mapping (SPM) with MATLAB 2018a (MATLAB, MathWorks, Natick, USA), in

255 accordance with Pataky et al., [50], using the source code available at http://www.spm1d.org/.

256 Differences between surfaces were examined using paired t-tests (SPM t). The alpha (α) level

257 for statistical significance for SPM was set at the 0.05 level.

258

259 Differences in discrete biomechanical parameters that could not be contrasted using SPM were

260 examined using Bayesian factors (BF) to explore the extent to which the data supported the

261 alternative (H_1) or null (H_0) hypotheses i.e. that there were or were no meaningful differences

262 between surface conditions for both males and females. Bayes factors were interpreted in

263 accordance with the recommendations of Jeffreys, [51], with values <1 indicating no evidence,

264 1-3 anecdotal evidence, 3-10 indicating substantial evidence, 10-30 strong evidence, 30-100

265 very strong evidence and >100 decisive evidence in support of H_1. In accordance with the

266 aforementioned recommendations, values >3 were considered sufficient evidence in support of

267 H_1. Finally, participants' subjective ratings were examined using Chi-squared (X^2) tests.

268 Discrete statistical tests were conducted using SPSS v25.0 (SPSS, USA).

269

270 **Results**

271 Knee joint kinetics, kinematics, muscle forces and GRFs contrasted using SPM are presented

272 in figures 3-4 and the discrete parameters are found in tables 1-2.

273

274   **@@@TABLE 1 NEAR HERE@@@**

275   **@@@TABLE 2 NEAR HERE@@@**

276

277   45° change of direction

278   *Statistical parametric mapping*

279   ACL force was shown to be significantly greater in the 2G surface from 15-25 and 70-100 %

280   of the stance phase (Figure 3D). The SPM analyses also showed that patellar tendon force,

281   patellofemoral force, patellofemoral stress and quadriceps force was significantly greater in the

282   indoor surface from 15-75 % of the stance phase (Figure 3E-H). In addition, hamstring forces

283   (Figure 3I) were significantly greater in the 2G surface from 0-50 and 85-95 % of the stance

284   phase and braking forces were significantly larger in the indoor surface from 0-5 and 20-80 %

285   of the stance phase (Figure 3K). Finally, the knee abduction moment was shown to be

286   significantly greater in the 2G surface from 10-95 % of the stance phase (Figure 3L).

287

288   **@@@FIGURE 3 NEAR HERE@@@**

289

290   *Discrete parameters*

291   For the translational coefficient of friction, quadriceps integral and peak rotational moment

292   there was decisive evidence in favour of these parameters being greater in the indoor condition.

293   There was also strong evidence in favour of the quadriceps force load rate being greater in the

294   indoor condition. Furthermore, for the hamstring integral and hamstring load rate there was

295   decisive evidence in favour of these parameters being greater in the 2G surface (Table 1).

296

297   Furthermore, for the patellar integral, patellofemoral force integral and patellofemoral stress

298   integral there was decisive evidence in favour of these parameters being greater in the indoor

299 condition. There was also substantial-strong evidence that the patellar load rate, patellofemoral

300 force load rate and patellofemoral stress load rates were larger in the indoor surface. Finally,

301 for the ACL integral and knee abduction moment integral there was decisive evidence in favour

302 of these parameters being greater in the 2G condition and strong evidence in favour of the knee

303 abduction moment load rate being larger in the 2G surface (Table 2).

304

305 180° change of direction

306 *Statistical parametric mapping*

307 ACL force was shown to be significantly greater in the 2G surface from 5-95 % of the stance

308 phase (Figure 4D). The SPM analyses also showed that patellar tendon force, patellofemoral

309 force, patellofemoral stress and quadriceps force was significantly greater in the indoor surface

310 from 20-30 % of the stance phase (Figure 4E-H). In addition, hamstring forces (Figure 4I) were

311 significantly greater in the 2G surface from 15-20 and 70-95 % of the stance phase and braking

312 forces were significantly larger indoor surface from 10-90 % of the stance phase (Figure 4K).

313 Finally, the knee abduction moment was shown to be significantly greater in the 2G surface

314 from 10-90 % of the stance phase (Figure 4L).

315

316   **@@@FIGURE 4 NEAR HERE@@@**

317

318 *Discrete parameters*

319 For the translational coefficient of friction, load rate braking and peak rotational moment there

320 was decisive evidence in favour of these parameters being greater in the indoor condition.

321 Furthermore, for the hamstring integral and there was decisive evidence in favour of these

322 parameters being greater in the 2G surface (Table 1).

323

324   Finally, for the ACL integral and knee abduction moment integral there was decisive evidence

325   in favour of these parameters being greater in the 2G condition and strong evidence in favour

326   of the knee abduction moment load rate being larger in the 2G surface (Table 2).

327

328   **Subjective ratings**

329   For the subjectively preferred surface, the chi-squared test was non-significant ($X^2 = 0.20$,

330   P>0.05) with nine participants reporting a preference for the indoor surface and eleven for the

331   2G surface. However, for the subjective ratings of surface traction the chi-squared test was

332   significant ($X^2 = 5.00$, P<0.05), with fifteen participants reporting that the indoor surface

333   provided more traction and five participants indicating the 2G surface.

334

335   **Discussion**

336   The aim of the current investigation was to examine the effects of 2G and indoor surfaces on

337   knee joint kinetics, kinematics, frictional and muscle force parameters during 45° and 180°

338   change of direction movements using SPM and Bayesian analyses. To the authors knowledge

339   this is the first investigation of this nature and may provide further important information

340   regarding the effects of different surfaces on the risk factors linked to the aetiology of knee

341   pathologies during functional athletic tasks.

342

343   This investigation importantly confirmed that the coefficient of friction and peak rotational

344   moment were greater in the indoor surface in relation to the 2G condition. This observation is

345   in agreement with the subjective ratings, which similarly showed that participants rated that

346   the indoor surface provided more traction. As there were notable differences in braking force

347   parameters observed using SPM and Bayesian analyses yet no differences in vertical GRFs, it

348   can be concluded that alterations in the coefficient of friction were mediated through alterations

349   in anterior-posterior GRFs. Importantly, frictional forces allow the resultant GRF vector to be

350 directed more effectively towards the intended direction, mediating enhanced linear

351 acceleration [52]. However, increased traction has also been linked to the aetiology of injury

352 [20, 53]. Nonetheless, owing to an enhanced coefficient of friction this observation suggests

353 that the indoor surface mediated an increased resistance to sliding compared to the 2G

354 condition.

355

356 Importantly the current investigation showed using SPM and Bayesian analyses that ACL

357 loading parameters were greater in the 2G condition, an interesting observation as hamstring

358 forces were larger and quadriceps forces were reduced in the 2G condition. As the quadriceps

359 load the ACL by mediating anterior tibial translation, and the hamstrings oppose tibial

360 translation and thus act to offload the ACL [20, 54], it could be expected that ACL forces would

361 be attenuated in the 2G surface. However, recent subject- specific musculoskeletal modelling

362 investigations have shown that the knee abduction moment is the inverse-dynamic mechanism

363 that most strongly governs the magnitude of ACL loading [55]. Indeed, the knee abduction

364 moment influences ACL loading by altering the tolerance of the ACL to anterior tibial

365 translation forces [55] and has been shown through a prospective in vivo investigation as the

366 biomechanical factor that most strongly predicts ACL injury [56]. It can therefore be

367 conjectured that the enhanced ACL loads experienced in the 2G surface conditions were

368 mediated via the enhanced knee abduction moments that were also revealed in this condition.

369 Importantly, the aetiology of ACL injuries in athletic populations is linked to excessive loading

370 of the ACL itself [16]. Therefore, owing to an enhanced ACL loading in the 2G surface, the

371 findings indicate that the specific 2G surface examined in this investigation may increase the

372 risk from ACL injury during sport specific change of direction movements compared to the

373 indoor surface.

374

375 In addition, the current study also showed using SPM and Bayesian analyses that both

376 patellofemoral and patellar tendon loading were larger in the indoor condition compared to the

377 2G surface. It is proposed that these observations were mediated through the increased

378 quadriceps forces in the 2G surface, as previous analyses have shown that quadriceps kinetics

379 strongly affect patellar tendon/ patellofemoral loading [27, 38]. This observation concurs with

380 the conclusions of Yu et al., [57] who showed that an enhanced coefficient of friction directly

381 increases the force of contraction from the quadriceps. This observation may be clinically

382 important as excessive patellar tendon/ patellofemoral joint loading are the mechanisms most

383 strongly linked to the aetiology of pain symptoms in active individuals [10, 11]. It can be

384 concluded on account of the enhanced tendon/ joint loading that the indoor surface examined

385 in the current investigation may increase the risk from chronic knee pathologies injury during

386 change of direction movements.

387

388 A potential limitation that should be acknowledged of the current investigation is that only male

389 athletes were examined. Female athletes have been shown to exhibit distinct external joint

390 moments [58], ACL loading [26, 46], lower extremity joint kinematics [58] and patellofemoral

391 joint stress [32] compared to male athletes. This suggests that further investigation into the

392 effects of different surfaces using a female sample is warranted before comprehensive

393 conclusions can be drawn.

394

395 In conclusion, although previous investigations have examined the biomechanical influence of

396 different surfaces, current knowledge regarding the effects of 2G and indoor surfaces on the is

397 biomechanics of change of direction movements is limited. As such the current investigation

398 contributes to biomechanical literature by providing a comprehensive examination of knee joint

399 kinetics, kinematics, frictional and muscle force parameters during 45° and 180° change of

direction movements. Importantly, this study showed using both SPM and Bayesian analyses that ACL loading parameters were greater in the 2G condition in relation to the indoor surface. Conversely, SPM and Bayesian analyses confirmed that patellofemoral/ patellar tendon loading alongside the coefficient of friction and peak rotational moment were larger in the indoor condition compared to the 2G surface. This study indicates that the indoor surface may improve change of direction performance owing to enhanced friction at the shoe-surface interface but augment the risk from patellar tendon/ patellofemoral injuries whereas the 2G condition may enhance the risk from ACL pathologies.

References

1. Schnohr P, O'Keefe JH, Marott JL, Lange P, Jensen GB (2015). Dose of jogging and long-term mortality: the Copenhagen City Heart Study. Journal of the American College of Cardiology 65: 411-419.

2. Blair SN, LaMonte MJ, Nichaman MZ (2004). The evolution of physical activity recommendations: how much is enough?. American Journal of Clinical Nutrition 79: 913-920.

3. Lauersen JB, Bertelsen DM, Andersen LB (2014). The effectiveness of exercise interventions to prevent sports injuries: a systematic review and meta-analysis of randomised controlled trials. British Journal of Sports Medicine 48: 871-877.

4. Conn JM, Annest JL, Gilchrist J (2003). Sports and recreation related injury episodes in the US population. Injury Prevention 9: 117–123.

5. Goldberg AS, Moroz L, Smith A, Ganley T. (2007). Injury surveillance in young athletes: a clinician's guide to sports injury literature. Sports Medicine 37: 265–278.

6. Marshall SW, Guskiewicz KM (2003). Sports and recreational injury: the hidden cost of a healthy lifestyle. Injury Prevention 9: 100-102.

425   7. Hootman JM, Dick R, Agel J (2007). Epidemiology of collegiate injuries for 15 sports:
426   summary and recommendations for injury prevention initiatives. Journal of Athletic
427   Training 42: 311-316.

428   8. John R, Dhillon MS, Syam K, Prabhakar S, Behera P, Singh H (2016). Epidemiological
429   profile of sports-related knee injuries in northern India: An observational study at a
430   tertiary care centre. Journal of Clinical Orthopaedics and Trauma 7: 207-211.

431   9. Smith BE, Selfe J, Thacker D, Hendrick P, Bateman M, Moffatt F, Logan P. (2018).
432   Incidence and prevalence of patellofemoral pain: A systematic review and meta-
433   analysis. PloS one 13: e0190892.

434   10. Crossley KM, Stefanik JJ, Selfe J, Collins NJ, Davis IS, Powers CM, McConnell J,
435   Vicenzino B, Bazett-Jones BM, Esculier J-F, Morrissey D, Callaghan MJ (2016).
436   Patellofemoral pain consensus statement from the 4th International Patellofemoral Pain
437   Research Retreat, Manchester. Part 1: Terminology, definitions, clinical examination,
438   natural history, patellofemoral osteoarthritis and patient-reported outcome measures.
439   British Journal of Sports Medicine 50: 839–843.

440   11. Rudavsky A, Cook J (2014). Physiotherapy management of patellar tendinopathy.
441   Journal of Physiotherapy 60: 122-129.

442   12. Lian ØB, Engebretsen L, Bahr R. (2005). Prevalence of jumper's knee among elite
443   athletes from different sports a cross-sectional study. American Journal of Sports
444   Medicine 33: 561-567.

445   13. Evans S, Shaginaw J, Bartolozzi A (2014). ACL reconstruction- it's all about timing.
446   International Journal of Sports Physical Therapy 9: 268-272.

447   14. Gottlob CA, Baker Jr CL, Pellissier JM, Colvin L (1999). Cost effectiveness of anterior
448   cruciate ligament reconstruction in young adults. Clinical Orthopaedics and Related
449   Research 367: 272-282.

450   15. Boden BP, Torg JS, Knowles SB, Hewett TE (2009). Video analysis of anterior cruciate

451   ligament injury abnormalities in hip and ankle kinematics. The American Journal of

452   Sports Medicine 37: 252-259.

453   16. Smith HC, Vacek P, Johnson RJ, Slauterbeck JR, Hashemi J, Shultz S, Beynnon BD

454   (2012). Risk factors for anterior cruciate ligament injury: a review of the literature—

455   part 1: neuromuscular and anatomic risk. Sports Health 4: 69-78.

456   17. Mears AC, Osei-Owusu P, Harland AR, Owen A, Roberts JR (2018). Perceived links

457   between playing surfaces and injury: A worldwide study of elite association football

458   players. Sports Medicine-Open 4: 40-44.

459   18. Fleming P (2011). Artificial turf systems for sport surfaces: current knowledge and

460   research needs. Proceedings of the Institution of Mechanical Engineers, Part P: Journal

461   of Sports Engineering and Technology 225: 43-63.

462   19. Thomas TD, Comfort P, Jones PA (2018). Comparison of change of direction speed

463   performance and asymmetries between team-sport athletes: Application of change of

464   direction deficit. Sports 6: 174-178.

465   20. Sinclair J, Stainton P (2017). Effects of specific and non-specific court footwear on

466   anterior cruciate ligament loading during a maximal change of direction manoeuvre.

467   Footwear Science 9: 161-167.

468   21. Thomson A, Whiteley R, Bleakley C (2015). Higher shoe-surface interaction is

469   associated with doubling of lower extremity injury risk in football codes: a systematic

470   review and meta-analysis. British Journal of Sports Medicine 49: 1245-1252.

471   22. Dragoo JL, Braun HJ, Harris AH (2013). The effect of playing surface on the incidence

472   of ACL injuries in National Collegiate Athletic Association American Football. The

473   Knee 20: 191-195.

474   23. Pataky TC, Robinson MA, Vanrenterghem J (2013). Vector field statistical analysis of

475   kinematic and force trajectories. Journal of Biomechanics 46: 2394-2401.

476   24. Pullenayegum EM, Thabane L (2009). Teaching Bayesian statistics in a health research

477   methodology program. Journal of Statistical Education 17: 21-23.

478   25. Cappozzo A, Catani F, Leardini A, Benedeti MG, Della CU (1995). Position and

479   orientation in space of bones during movement: Anatomical frame definition and

480   determination. Clinical Biomechanics 10: 171-178.

481   26. Sinclair J, Brooks D, Stainton P (2019). Sex differences in ACL loading and strain

482   during typical athletic movements: a musculoskeletal simulation analysis. European

483   Journal of Applied Physiology, 119, 713-721.

484   27. van Eijden TM, Kouwenhoven E, Verburg J, Weijs WA (1986). A mathematical model

485   of the patellofemoral joint. Journal of Biomechanics19: 219–229.

486   28. Willson JD, Ratcliff OM, Meardon SA, Willy RW (2015). Influence of step length and

487   landing pattern on patellofemoral joint kinetics during running. Scandinavian Journal

488   of Medicine & Science in Sports 25: 736-743.

489   29. Sinclair J (2014). Effects of barefoot and barefoot inspired footwear on knee and ankle

490   loading during running. Clinical biomechanics 29: 395-399.

491   30. Sinclair J (2018). Mechanical effects of medial and lateral wedged orthoses during

492   running. Physical Therapy in Sport 32: 48-53.

493   31. Sinclair J, Butters B, Brooks D, Stainton P. (2018). Effects of a prophylactic knee

494   bracing on patellofemoral loading during cycling. Sport Sciences for Health 14: 645-

495   654.

496   32. Sinclair J, Selfe J. (2015). Sex differences in knee loading in recreational runners.

497   Journal of Biomechanics 48: 2171-2175.

498   33. Sinclair J, Bottoms L (2015). Gender differences in patellofemoral load during the epee
499   fencing lunge. Research in Sports Medicine 23: 51-58.

500   34. Heino JB, Powers CM (2002). Patellofemoral stress during walking in persons with and
501   without patellofemoral pain. Medicine & Science in Sports & Exercise 34: 1582-1593.

502   35. DeVita P, Hortobagyi T (2001). Functional knee brace alters predicted knee muscle and
503   joint forces in people with ACL reconstruction during walking. Journal of Applied
504   Biomechanics 17: 297–311.

505   36. Spoor CW, van Leeuwen JL (1992). Knee muscle moment arms from MRI and from
506   tendon travel. Journal of Biomechanics 25: 201–206.

507   37. Besier TF, Draper CE, Gold GE, Beaupre GS, Delp SL (2005). Patellofemoral joint
508   contact area increases with knee flexion and weight-bearing. Journal of Orthopaedic
509   Research 23: 345–350.

510   38. Janssen I, Steele JR, Munro BJ, Brown NA (2013). Predicting the patellar tendon force
511   generated when landing from a jump. Medicine & Science in Sports & Exercise. 45:
512   927-934.

513   39. Sinclair J, Richards JD, Taylor PJ. (2018). Effects of prophylactic knee bracing on
514   patellar tendon loading parameters during functional sports tasks in recreational
515   athletes. Sport Sciences for Health 14: 151-160.

516   40. Sinclair J, Taylor PJ (2015). Sex variation in patellar tendon kinetics during running.
517   Human Movement 16: 60-63.

518   41. Richards J, Selfe J, Sinclair J, May K, Thomas G. (2016). The effect of different decline
519   angles on the biomechanics of double limb squats and the implications to clinical and
520   training practice. Journal of Human Kinetics 52: 125-138.

521   42. Sinclair JK, Shore H, Atkins S, Hobbs SJ (2015). Side to side differences in patellar
522   tendon kinetics of the support limb during maximal instep soccer kicking. Baltic Journal
523   of Health and Physical Activity 7: 20-24.

524   43. Herzog W, Read LJ (1993). Lines of action and moment arms of the major force-
525   carrying structures crossing the human knee joint. Journal of Anatomy 182: 213-230.

526   44. Dai B, Yu B. (2012). Estimating ACL force from lower extremity kinematics and
527   kinetics. 36th annual meeting of the American Society of Biomechanics Gainesville,
528   Florida, 253-254.

529   45. Sinclair J, Taylor PJ (2019). Effects of a Prophylactic Knee Sleeve on Anterior Cruciate
530   Ligament Loading During Sport-Specific Movements. Journal of sport Rehabilitation
531   28: 1-7.

532   46. Sinclair J, Bottoms L. (2019). Gender specific ACL loading patterns during the fencing
533   lunge: Implications for ACL injury risk. Science & Sports, 34, 31-35.

534   47. Cerulli G, Benoit DL, Lamontagne M, Caraffa A, Liti A. (2003). In vivo anterior
535   cruciate ligament strain behaviour during a rapid deceleration movement: case report.
536   Knee Surgery, Sports Traumatology, Arthroscopy 11: 307–311.

537   48. Krosshaug T, Slauterbeck JR, Engebretsen L, Bahr R (2007). Biomechanical analysis
538   of anterior cruciate ligament injury mechanisms: three-dimensional motion
539   reconstruction from video sequences. Scandinavian Journal of Medicine and Science in
540   Sports 17: 508–519.

541   49. Holden JP, Cavanagh PR (1991). The free moment of ground reaction in distance
542   running and its changes with pronation. Journal of Biomechanics 24: 891-389.

543   50. Pataky TC, Robinson MA, Vanrenterghem J (2016). Region-of-interest analyses of
544   one-dimensional biomechanical trajectories: bridging 0D and 1D theory, augmenting
545   statistical power. Peer J 4: 2652-2664

546   51. Jeffreys H. Theory of probability (3rd Ed.). 1961. Oxford, UK: Oxford University

547   Press.

548   52. Luo G, Stefanyshyn D (2011). Identification of critical traction values for maximum

549   athletic performance. Footwear Science 3: 127-138.

550   53. Nigg BM, Segesser B (1988). The influence of playing surfaces on the load on the

551   locomotor system and on football and tennis injuries. Sports Medicine 5: 375-385.

552   54. Withrow TJ, Huston LJ, Wojtys EM, Ashton-Miller JA (2008). Effect of varying

553   hamstring tension on anterior cruciate ligament strain during in vitro impulsive knee

554   flexion and compression loading. The Journal of Bone and Joint Surgery. American

555   volume 90: 815-823.

556   55. Navacchia A, Bates NA, Schilaty ND, Krych AJ, Hewett TE (2019). Knee abduction

557   and internal rotation moments increase ACL force during landing through the posterior

558   slope of the tibia. Journal of Orthopaedic Research 37: 1730-1742.

559   56. Hewett TE, Myer GD, Ford KR (2005). Biomechanical measures of neuromuscular

560   control and valgus loading of the knee predict anterior cruciate ligament injury risk in

561   female athletes. American Journal of Sports Medicine 33: 492-501.

562   57. Yu B, Lin CF, Garrett WE (2006). Lower extremity biomechanics during the landing

563   of a stop-jump task. Clinical Biomechanics 21: 297-305.

564   58. Ferber R, Davis IM, Williams DS (2003). Gender differences in lower extremity

565   mechanics during running. Clinical Biomechanics18: 350-357.

566

567   **Figure labels**

568   Figure 1: Experimental marker configuration.

569   Figure 2: Pelvic, thigh, tibial and foot segments, with segment co-ordinate system axes

570   (X=sagittal plane; Y=coronal plane and Z=transverse anatomical planes).

571   Figure 3: Kinetic and kinematic parameters as a function of surface for the 45° change of

572   direction movement (Black = 2G & Red = Indoor).

573   Figure 4: Kinetic and kinematic parameters as a function of surface for the 180° change of

574   direction movement (Black = 2G & Red = Indoor). (Black = 2G & Red = Indoor).

Table 1: Frictional and muscle force parameters (mean±SD) as a function of the experimental movement and surface conditions.
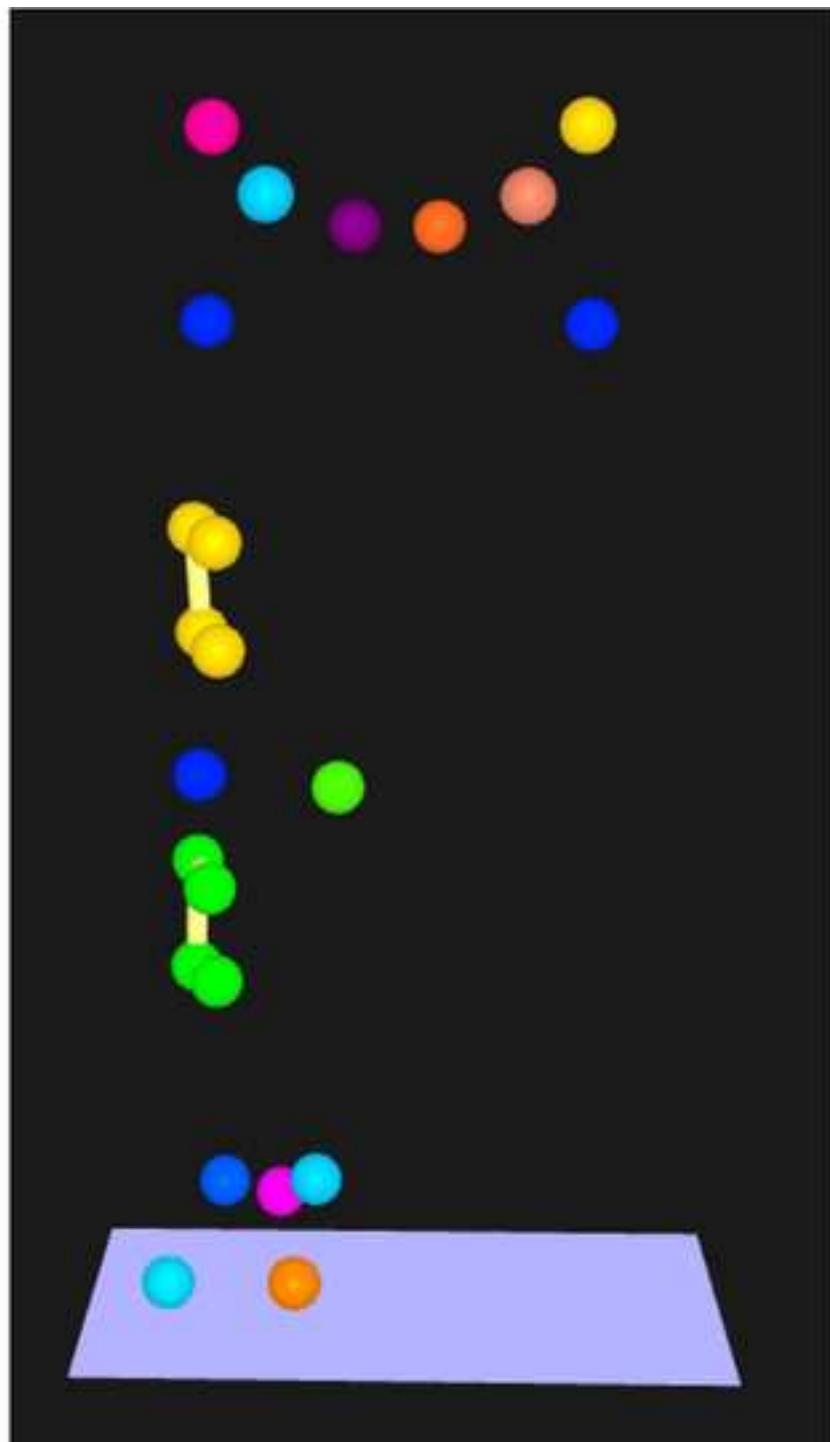
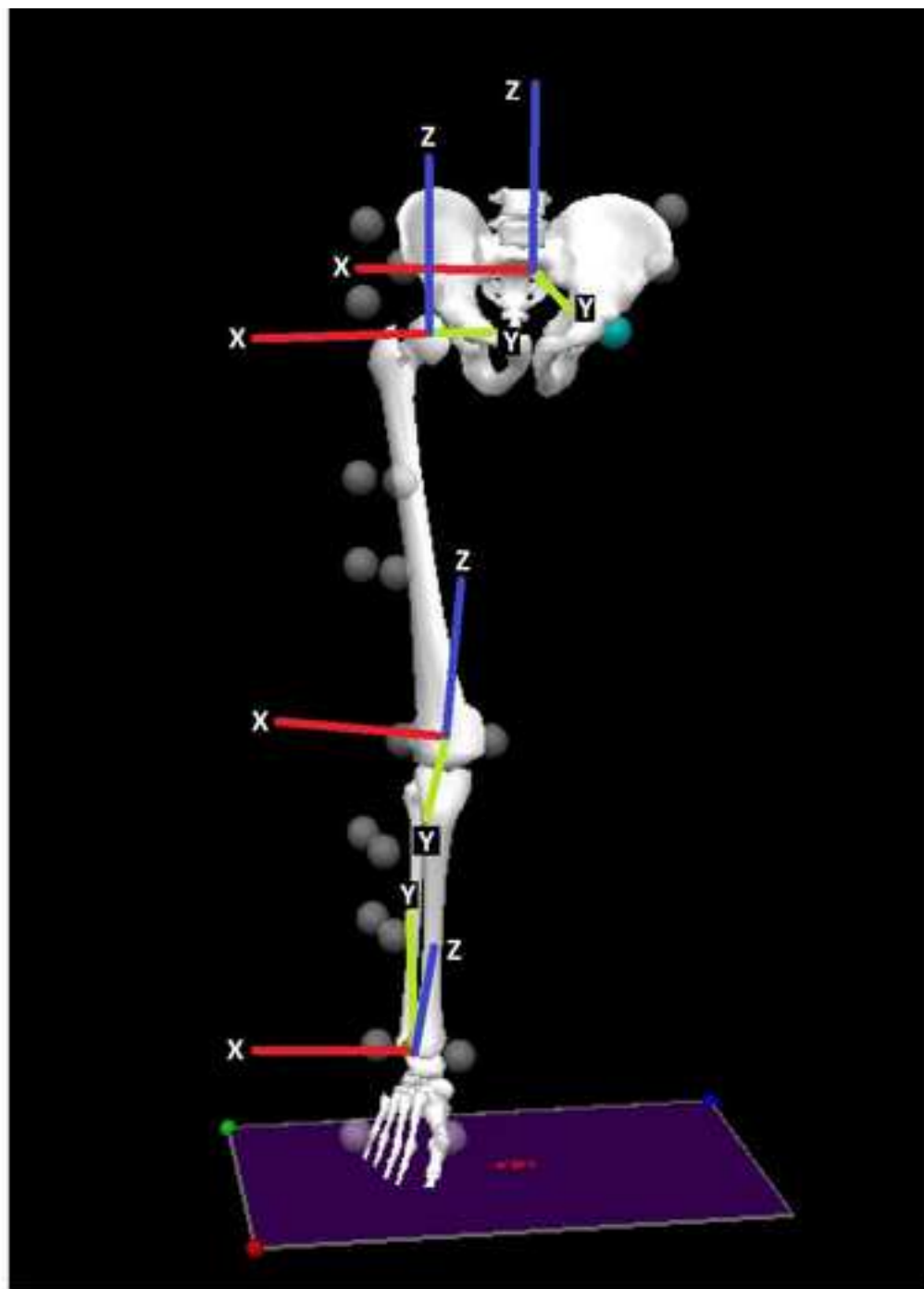|	45°				Bayes factor	180°				Bayes factor
	2G		Indoor			2G		Indoor		
	Mean	SD	Mean	SD		Mean	SD	Mean	SD	
Translational coefficient of friction (μ)	0.18	0.06	0.43	0.22	*1955.58*	0.31	0.04	0.68	0.12	*3790913148*
Load rate vertical GRF (N/s)	257380.63	145386.71	213585.60	166827.07	*0.98*	91427.29	44069.39	90583.25	41206.03	*0.22*
Load rate braking (N/s)	105688.64	68411.50	113758.53	88612.97	*0.25*	41308.94	25208.46	57044.95	25020.64	*256.22*
Quadriceps integral (N·s)	519.07	205.05	799.63	327.81	*2122.58*	1232.12	617.79	1356.81	441.27	*0.37*
Quadriceps force load rate (N/s)	294927.37	89591.22	400281.59	184320.64	*13.81*	182912.39	69116.06	215434.26	73472.04	*1.45*
Hamstring integral (N·s)	418.28	245.20	172.63	73.43	*700.97*	776.07	357.91	485.60	299.91	*114.61*
Hamstring load rate (N/s)	451553.79	262740.43	355276.42	218654.34	*231.17*	163808.19	68777.20	160825.39	62659.50	*0.22*
Peak rotational moment (Nm)	10.71	4.95	17.23	7.73	*843.31*	6.35	2.23	19.86	7.66	*405689*

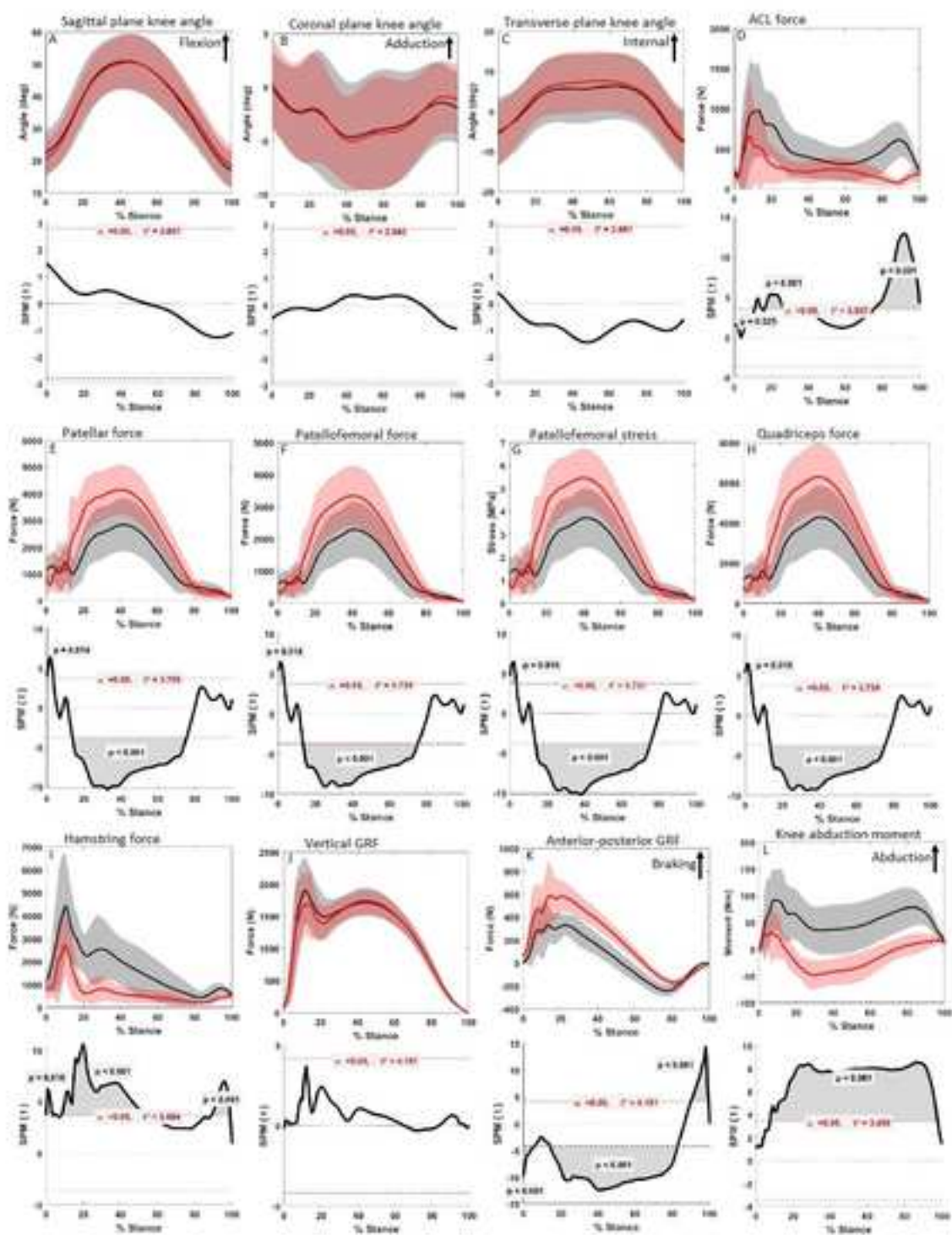*Notes: Bold Bayes factors indicate values >3, i.e. at least substantial evidence in support of H_1.*

Table 2: Knee joint loading parameters (mean±SD) as a function of the experimental movement and surface conditions.
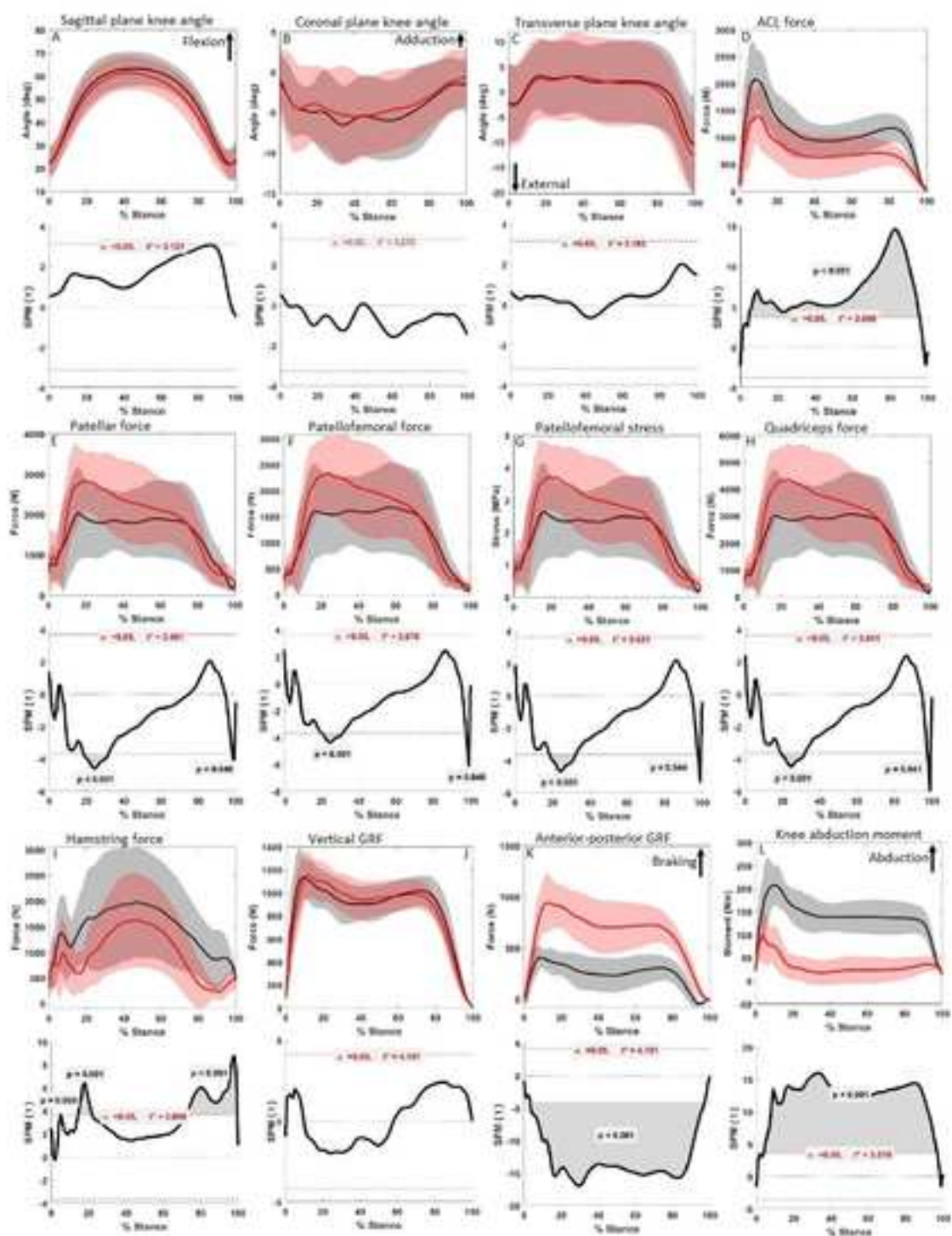
	45°				*Bayes factor*	180°				*Bayes factor*
	2G		**Indoor**			**2G**		**Indoor**		
	Mean	*SD*	**Mean**	*SD*		**Mean**	*SD*	**Mean**	*SD*	
ACL integral (N·s)	120.04	60.17	60.12	29.30	***184.85***	569.31	157.10	336.58	103.41	***9738740***
ACL load rate (N/s)	130835.62	98683.33	101851.46	105484.49	*2.92*	151352.59	79027.39	136652.04	85639.02	*0.39*
Patellar integral (N·s)	368.81	126.16	556.94	201.50	***2656.68***	808.91	390.88	897.91	288.06	*0.43*
Patellar load rate (N/s)	245518.44	75613.85	346615.91	189618.36	***5.31***	146763.38	51995.62	181376.58	77201.91	*1.63*
Patellofemoral force integral (N·s)	272.08	112.17	419.84	178.95	***1545.97***	661.33	336.80	724.64	239.71	*0.35*
Patellofemoral force load rate (N/s)	149456.58	46041.49	201387.17	90435.87	***16.12***	93317.93	35693.47	109493.77	36830.67	*1.49*
Patellofemoral stress integral (MPa·s)	0.47	0.16	0.72	0.27	***3993.67***	1.04	0.51	1.16	0.37	*0.45*
Patellofemoral stress load rate (MPa/s)	295.30	88.25	409.72	207.64	***7.63***	179.01	64.70	216.02	81.14	*1.62*
Knee abduction moment integral (Nm·s)	14.28	8.50	-3.65	3.34	***4436895***	73.01	27.13	13.71	13.44	***81161120***
Knee abduction moment load rate (Nm/s)	13815.90	7780.47	9684.01	7381.42	***28.85***	15426.31	7695.95	12834.24	10020.31	*0.55*

Notes: Bold Bayes factors indicate values >3, i.e. at least substantial evidence in support of H_1.

Reviewer #1:

COMMENTS TO AUTHORS

Dear Authors,

I believe the topic of your article is interesting.

However, there are some points I would clarify, before publication.

I approve the publication of this paper after minor revision.

I have the following detailed comments:

TITLE

Ok

ABSTRACT

Well written

INTRODUCTION

The introduction provides adequate background.

METHODS

OK

RESULTS

OK

DISCUSSION/ CONCLUSION

It is recommend to reinforce the motivations and criteria that led you to the conclusions through this methodology and discuss why the applied method is appropriate.

RESPONSE: The mechanisms responsible for each conclusion is now added to each paragraph or relevance in the discussion.

With regards to the modelling methods with great respect, we feel that the efficacy of each model is already described in the methods section as part of the alterations included in the previous revision.

REFERENCES

Ok.

FIGURES

OK