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1 **The sensitivity of joint kinematics and kinetics to marker placement**
2 **during a change of direction task**

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21 **The sensitivity of joint kinematics and kinetics to marker placement**
22 **during a change of direction task.**

23 **Abstract**

24 The conventional gait model (CGM) refers to several closely related biomechanical
25 models used in the objective analysis of human motion. Their use has become
26 popular in the analysis of change of direction tasks to inform best practice in the
27 prevention and rehabilitation of anterior cruciate ligament injury. As externally-placed
28 markers define segment axes origins and orientations, kinematic and kinetic outputs
29 from the CGM are sensitive to marker placement. The aim of this investigation was
30 to quantify the sensitivity of lower extremity kinematics and knee moments to
31 systematic differences in marker placement across the stance phase of a change of
32 direction task. Systematic anterior/posterior displacements were applied to the lateral
33 thigh, femoral epicondyle and tibia markers in software. One-dimensional statistical
34 parametric mapping was used to determine the effect of marker placement across
35 the entire stance phase of a 90° change of direction task. Marker placement error
36 within previously reported inter-tester variability ranges caused significant differences
37 in knee abduction moment, hip rotation angle, knee rotation angle, ankle rotation
38 angle and ankle abduction angle across various periods of stance. Discrete
39 measures of these variables have been associated with increased frontal plane knee
40 loading during change of direction, considered a key mechanism of anterior cruciate
41 ligament injury. Systematic differences in marker placement may lead to incorrect
42 group statistical inferences in such discrete measures.

43

44

45 **Introduction**

46 The conventional gait model (CGM) refers to several closely related biomechanical
47 models, the data from which are used to analyse human motion, inform clinical
48 decision making and evaluate rehabilitation interventions (Baker et al. 2017). Such
49 models provide an objective record of kinematic and kinetic metrics during
50 movement. Originally developed for and implemented in clinical gait analyses, the
51 CGM's application has been extended to a variety of movements, including a range
52 of change of direction (CoD) tasks (Franklyn-Miller et al. 2017; King, Richter,
53 Franklyn-Miller, Daniels, Wadey, Jackson, et al. 2018; B. M. Marshall et al. 2014;
54 McLean, Huang, and Van Den Bogert 2005; O'Malley et al. 2018; Sigward and
55 Powers 2007).

56 CoD is the most common mechanism of non-contact anterior cruciate ligament
57 (ACL) rupture, a serious musculoskeletal injury normally requiring surgical
58 intervention (Kvist 2004). The CGM has been utilised in the analysis of CoD to
59 inform best practice in the prevention and rehabilitation of ACL injury (King, Richter,
60 Franklyn-Miller, Daniels, Wadey, Jackson, et al. 2018; McLean, Huang, and Van Den
61 Bogert 2005; Sigward and Powers 2007). Kinematic variables at the hip, knee and
62 ankle have been associated with increased frontal plane knee loading during CoD,
63 considered a key risk factor for injury (Hewett et al. 2005; McLean, Huang, and Van
64 Den Bogert 2005; Sigward and Powers 2007).

65 Accurate measures of these variables rely on the correct definition of body segment
66 axes origins and orientations (Kadaba et al. 1989). In the Plug-in-Gait (PiG) model
67 (Vicon, Oxford Metrics, London, UK), a widely used implementation of the CGM,
68 retroreflective markers placed externally on a series of anatomical landmarks define

69 segment origins and orientations. Variation in marker placement is cited as the
70 primary factor in the low reliability indices reported for many kinematic and kinetic
71 variables (Alenezi et al. 2016; Gorton, Hebert, and Gannotti 2009; McGinley et al.
72 2009).

73 Inter-tester variability in anatomical landmark location, and subsequently marker
74 placement, makes inferring ACL injury mechanisms based on data collected in
75 different laboratories and by different practitioners challenging. The range of inter-
76 tester variability in anatomical landmark location for marker positions has been
77 reported as 12 – 25 mm (Della Croce, Cappozzo, and Kerrigan 1999). Given their
78 roles in defining the origins and orientations of the femur and shank segments, the
79 lateral thigh (THI), lateral femoral epicondyle (KNEE) and lateral tibia (TIB) markers
80 have the largest effect on model outputs (Kadaba, Ramakrishnan, and Wooten
81 1989). The deterministic nature of the model indicates that variation in the
82 anterior/posterior positions of these markers will alter joint kinematics and kinetics at
83 the hip, knee and ankle (Kadaba, Ramakrishnan, and Wooten 1989).

84 Experimental studies confirm the sensitivity of joint kinematics, particularly frontal
85 and transverse plane kinematics, to marker placement error during walking (Baker,
86 Finney, and Orr 1999b; Ferrari et al. 2008; Groen et al. 2012; Kadaba et al. 1989;
87 Szcserbik and Kalinowska 2014). Simulated displacements in THI marker position
88 cause large errors in transverse plane hip and frontal plane knee kinematics, both of
89 which have been associated with increased frontal plane knee loading during CoD
90 (Baker, Finney, and Orr 1999b; McLean, Huang, and Van Den Bogert 2005; Sigward
91 and Powers 2007). Errors in frontal plane knee kinematics vary non-uniformly
92 throughout the gait cycle, demonstrating analysis of the entire gait cycle may be
93 required to fully understand the effect of marker placement on joint kinematics.

94 Calculated joint moments of force are also affected by marker placement. Changing
95 the positions of the THI, KNEE and TIB markers alters the locations of the calculated
96 knee (KJC) and ankle joint centres (AJC), affecting the length of the moment arm
97 used to calculate the joint moment. Simulated displacements in joint centre positions
98 demonstrate this, with 10 mm anterior displacements causing significant differences
99 in net knee moments during walking (Holden and Stanhope 1998; Stagni et al.
100 2000).

101 The specific sensitivity of kinematic and kinetic variables to systematic differences in
102 marker placement remains unclear. The effect of marker placement will vary
103 depending on the variable being reported, the marker in question, the magnitude of
104 displacement and the phase of the movement being analysed. To reliably make
105 inferences related to ACL injury from data collected in different laboratories and by
106 different practitioners, we must establish the sensitivity of lower extremity kinematics
107 and knee moments to systematic differences in marker placement. The aim of this
108 investigation was to determine the sensitivity of joint kinematics at the hip, knee and
109 ankle, as well as knee moments, to systematic displacements in the positions of the
110 THI, KNEE and TIB markers across the stance phase of a CoD task.

111 **Methods**

112 Participants

113 An *a priori* power analysis (G*Power, version 3.1.9.2, Universität Düsseldorf,
114 Germany), based on previously published data (Alenezi et al. 2016), indicated that a
115 sample size of 42 participants was required to achieve 80% statistical power with an
116 alpha level of 0.05. Fifty eligible participants (mean \pm SD: 24.8 \pm 4.8 years, 180 \pm 6

117 cm and 84 ± 15.3 kg) were consecutively recruited from the caseload of two
118 orthopaedic surgeons based in the Sports Surgery Clinic, Dublin, Ireland.
119 Inclusion criteria for participation were: male, aged 18 – 35, undergone primary
120 ACLR 34 – 43 weeks (mean \pm SD: 35.7 ± 1.2 weeks) prior to testing, participation in
121 multi-directional field-based sport prior to ACL injury and intention to return to the
122 same level of participation following rehabilitation. The study received ethical
123 approval from the University of Roehampton, London (LSC 15/122) and the Sports
124 Surgery Clinical Hospital Ethics committee (25AFM010). Participants gave informed,
125 written consent prior to participation in the study.

126 Data Collection

127 Testing took place in a biomechanics laboratory, using a ten-camera motion analysis
128 system (200 Hz; Bonita-B10, Vicon, UK), synchronized (Vicon Nexus 2.7) with two
129 force platforms (1000 Hz BP400600, AMTI, USA) recording the positions of 28
130 reflective markers (14 mm diameter). Markers were secured to the participant's shoe
131 or skin using tape at bony landmarks on the lower limbs, pelvis and trunk according
132 to the PiG marker set (B. M. Marshall et al. 2014).

133 Prior to data collection, participants undertook a standardised warm-up comprising of
134 a 2-minute jog, 5 bodyweight squats, 2 submaximal and 3 maximal
135 countermovement jumps. A static trial was captured as a reference for the dynamic
136 trials. Each participant completed a pre-planned 90° CoD task. The CoD task
137 followed a wider testing battery that formed part of a larger, ongoing study, in which
138 participants also completed a range of double and single leg jump exercises. The
139 CoD task involved the participants running maximally towards the force platforms
140 then planting their outside foot on the force platform to cut left or right, i.e. planting

141 their left foot to cut to the right. Three valid, maximal effort trials were collected on
142 both the non-operated and operated limb. A full description of the testing protocol is
143 given in King et al. (2018).

144 Data Processing

145 Trials in which the participant planted their operated limb on the force platform to
146 complete the CoD task were used for further analysis. Marker trajectory and force
147 data were low-pass filtered using a fourth-order Butterworth filter (cut-off frequency
148 15 Hz) (Kristianslund, Krosshaug, and Bogert 2012). Systematic displacements were
149 then applied in software to the positions of the THI, KNEE and TIB markers. One
150 marker position displacement was applied at a time along the corresponding
151 segment x-axis using

$$152 \quad X_k' = T.X_k$$

153 where X_k' are the new, displaced marker coordinates within the segment coordinate
154 system, T is the translational matrix and X_k are the original marker coordinates within
155 the segment coordinate system (Fig 1). Displacements were applied to marker
156 positions in 5 mm increments, to 20 mm anterior and 20 mm posterior from their
157 original positions, resulting in 8 displacement conditions for each marker. Data
158 processing created three separate data sets: A, B and C. Each data set contained
159 displacements of a single marker and were identical except for the position of the
160 corresponding marker.

161 Stance phase was identified for each trial from when vertical ground reaction force
162 passed above and below 20 N. Tri-planar joint angles at the hip, knee and ankle, as
163 well as tri-planar knee moments were extracted during stance phase for each trial.

164 Kinematic and kinetic signals were time normalised to 101 data points and the mean
165 of each participant's three trials was used for further analysis.

166 Sensitivity Analysis

167 One-dimensional statistical parametric mapping (SPM) was used to analyse the
168 effect of marker placement across the entire stance phase of the CoD task (Pataky
169 2010, 2014; Pataky, Robinson, and Vanrenterghem 2013). Our analysis aimed to
170 simulate a scenario in which we were testing for between group differences in
171 groups which were identical except for the position of the corresponding marker. This
172 would allow us to identify the minimum systematic differences in marker placement
173 required to result in incorrect statistical inferences when making between group
174 comparisons in each variable. For clarity, we will use the example of one data set,
175 data set A, as the process was repeated identically for data sets B and C. Following
176 data processing, nine signals for each variable for each participant were contained in
177 data set A. These corresponded to the original unaltered trial, as well as each of the
178 THI marker displacement conditions (Fig 3).

179

180 Each variable in data set A was submitted to a 1D independent samples SPM t-test
181 between the unaltered condition and each of the displacement conditions. This
182 process produced 8 SPM $\{t\}$ curves for each variable, one for each THI marker
183 displacement condition (Fig 4). The significance of each SPM $\{t\}$ curve was
184 determined topologically using random field theory ($\alpha < 0.05$) (Pataky,
185 Vanrenterghem, and Robinson 2015). Phases of the SPM $\{t\}$ curve above the critical-
186 t threshold were identified as significantly affected by the corresponding marker
187 displacement. To aid in interpretation of results, SPM $\{t\}$ curves were plotted using

188 image inference surface plots (Fig. 5). A variable's "sensitivity" to marker placement
189 was determined by the minimum marker displacement required to cause significant
190 differences, with more sensitive variables significantly affected by smaller marker
191 displacements across larger periods of stance phase.

192 As we experimentally created the difference between conditions by displacing each
193 marker in a fixed direction from its original position, the changes to outcome
194 variables will be unidirectional and predictable in nature. For example, an anterior
195 displacement of the THI marker will always result in a more internally-rotated
196 calculated position of the thigh segment. The test statistic produced following
197 comparisons between the unaltered condition and each displacement condition is
198 therefore a function of sample size and effect size, meaning that the likelihood of
199 finding a statistically significant differences between conditions is increased at larger
200 sample sizes. In acknowledgment of this, we included sample size as an extra
201 degree of freedom in our analysis. We chose sample sizes of $n = 10$, $n = 25$ and $n =$
202 50 , as these represent the low, mid and upper ranges of sample sizes typically used
203 in biomechanical studies (Besier, Lloyd, and Ackland 2003; Ithurburn et al. 2017;
204 Sankey et al. 2015; Wen et al. 2018). The sensitivity analysis procedure outlined
205 above was repeated for each variable in data sets A, B and C, at each sample size,
206 resulting in a total of nine sensitivity analyses.

207 **Results**

208 The results of the sensitivity analyses for the THI, KNEE and TIB markers are
209 presented in Figures 6, 7 and 8 respectively. See supplementary material –
210 Appendix A, for individual sensitivity analyses for each variable. As sample size
211 increased, the magnitude of the marker displacement required to cause significant

212 differences in each variable decreased, and/or the cumulative percentage of stance
213 phase significantly affected by marker displacements increased.

214 Thigh Marker

215 No variables were significantly affected by 5mm THI marker displacements. Four
216 variables were significantly affected by displacements of 10 mm and greater across
217 periods of early, mid and late stance (Fig 5B, 6C). These variables were hip rotation
218 angle, knee abduction angle, ankle abduction angle and ankle rotation angle. Of
219 these, hip rotation and knee abduction angles were most sensitive to THI marker
220 placement, with 10 mm displacements causing significant differences across the
221 entire stance phase at $n = 50$ (Fig 5C). At $n = 10$, only hip rotation and knee
222 abduction angles were significantly affected by THI marker displacements of any
223 magnitude. The sensitivity of these variables increased as sample size increased,
224 while at $n = 25$ and $n = 50$, ankle abduction and rotation angles were also
225 significantly affected (Fig 5B, 5C).

226 Knee Marker

227 No variables were significantly affected by 5 mm KNEE marker displacements (Fig
228 6). Eight variables were significantly affected by KNEE marker displacements of 10
229 mm and above (Fig 6C). These were hip rotation angle, knee flexion angle, knee
230 rotation angle, ankle plantar-flexion angle, ankle abduction angle, knee flexor
231 moment and knee abduction moment (Fig 6B, 6C). Of these, ankle abduction and
232 rotation angles were most sensitive to KNEE marker displacements, with 10 mm
233 displacements causing significant differences across the first and last 20% of stance
234 (Fig 6C). At $n = 10$, no variables were significantly affected by KNEE marker
235 displacements of any magnitudes. At $n = 25$, ankle plantar-flexion, ankle abduction,

236 ankle rotation, knee flexor moment and knee abduction moment were significantly
237 affected (Fig 6B), while at n = 50, hip rotation, knee flexion, knee abduction and knee
238 rotation angles were also significantly affected (Fig 6C).

239

240 Tibia Marker

241 5 mm TIB marker displacements significantly affected three kinematic variables (Fig
242 7C). These were, knee rotation angle, ankle abduction angle and ankle rotation
243 angle. Displacements of 10 mm and above also significantly affected ankle plantar-
244 flexion angle, knee flexor moment and knee abduction moment (Fig 7B, 7C). Knee
245 rotation angle was the most sensitive variable to TIB marker displacements, and the
246 only variable to be significantly affected across the entire stance phase by any 5 mm
247 marker displacements (Fig 7C). At n = 10, knee rotation angle, ankle abduction
248 angle, ankle rotation angle and knee abduction moment were significantly affected
249 by TIB marker displacements (Fig 7C). The sensitivity of these variables increased
250 as sample size increased, while ankle plantar-flexion angle and knee abduction
251 moment were also significantly affected at n = 25 and n = 50 (Fig 7B, 7C).

252

253 **Discussion**

254 Inter-tester variability in the anterior/posterior positions of the anatomical landmarks
255 used to define the positions of the THI, KNEE and TIB markers is reported as
256 ranging between 9.3 – 12.5 mm (Della Croce, Cappozzo, and Kerrigan 1999).
257 Several variables previously associated with ACL injury risk and rehabilitation status
258 were significantly affected by marker displacements within, or bordering on, reported
259 inter-tester variability ranges. These were hip rotation angle, knee abduction angle,

260 ankle rotation angle and knee abduction moment (Dempsey et al. 2007; McLean,
261 Huang, and Van Den Bogert 2005; Sigward and Powers 2007).

262 Frontal and transverse plane kinematics were most sensitive to marker placement in
263 each marker condition and at every sample size. This is unsurprising given the
264 known limitations of the CGM in assessing frontal and transverse plane kinematics
265 (Baker, Finney, and Orr 1999a; Kadaba, Ramakrishnan, and Wooten 1989).

266 Changes in the anterior/posterior positions of the THI, KNEE and TIB markers
267 causes misalignment of the primary and secondary axis of the femur and shank
268 segments. These alterations create a rotational offset, while also resulting in cross-
269 talk between segment axes. This manifests as error in angles calculated in all three
270 planes, and is most pronounced in the frontal and transverse plane kinematics
271 (Baker, Finney, and Orr 1999b). Previous studies using descriptive statistics
272 (Szczerbik and Kalinowska 2011), root mean square differences (Groen et al. 2012)
273 and qualitative assessments (Kadaba et al. 1989) to examine the effect of marker
274 placement on joint kinematics during walking report similar findings.

275 Our findings build on those from previous work and demonstrate the minimum
276 systematic differences in marker placement required to cause statistically significant
277 differences in each variable at three different sample sizes. Utilising a continuous
278 statistical analysis method (SPM) allowed us to identify the specific phases of each
279 kinematic and kinetic signal significantly affected by marker displacements.

280 Statistically significant differences first appeared in many outcome variables across
281 the first and last 20% of stance, indicating these phases are most sensitive to marker
282 placement (Fig 5A, 6B, 7A). As non-contact ACL injuries are believed to occur within
283 the first 20% of stance, discrete kinematic and kinetic measures from this period are
284 regularly reported (Pollard, Sigward, and Powers 2007a; Sigward and Powers 2007;

285 Stearns and Pollard 2013). Increased hip internal rotation, knee abduction and ankle
286 external rotation at initial contact of CoD have been associated with higher peak
287 knee abduction moments (Dempsey et al. 2007; McLean, Huang, and Van Den
288 Bogert 2005; Sigward and Powers 2007). Frontal plane knee loading is considered a
289 key risk factor for ACL injury (Hewett et al. 2005). These findings have thus led to the
290 clinical development of ACL prevention and rehabilitation programs aiming to
291 minimise frontal plane knee loading (Distefano et al. 2011).

292 Statistical significance is often used to draw clinical inferences in ACL research
293 (Dempsey et al. 2007; Ford et al. 2005; King, Richter, Franklyn-Miller, Daniels,
294 Wadey, Jackson, et al. 2018; Sigward and Powers 2007; Stearns and Pollard 2013).
295 Previous work has reported statistically significant differences in kinematics and
296 kinetics with respect to gender (Ford et al. 2005), limbs (King, Richter, Franklyn-
297 Miller, Daniels, Wadey, Jackson, et al. 2018) and injured/uninjured groups (Stearns
298 and Pollard 2013) and postulated that these differences may highlight variables of
299 interest in rehabilitation and injury prevention. It should be noted that statistical
300 significance is less relevant than the actual magnitude of differences between groups
301 and how such differences would affect clinical inferences/recommendations. Relative
302 to previously published differences, our findings demonstrate magnitudes
303 approximating or exceeding those reported between groups/conditions (Ford et al.
304 2005; King, Richter, Franklyn-Miller, Daniels, Wadey, Jackson, et al. 2018; Pollard,
305 Sigward, and Powers 2007b; Stearns and Pollard 2013). For example, statistically
306 significant differences in hip rotation angle (5.1°), knee abduction angle (2°) and
307 knee abduction moment (0.21, 0.53 and 1 Nm/kg) during CoD tasks have been
308 reported previously and hypothesised to present clinically relevant differences
309 related to ACL injury (McLean, Huang, and Van Den Bogert 2005; Sigward and

310 Powers 2007; Stearns and Pollard 2013). Within our data, at $n = 50$ 10 mm THI
311 marker displacements caused significant differences in hip rotation and knee
312 abduction angle with a mean difference of 3.62° and 2.77° respectively, while 10 mm
313 TIB marker displacements caused significant differences in knee abduction moment
314 with a mean difference of 3.22 Nm/kg (see supplementary material – Appendix A).

315

316 Several limitations can be ascribed to the current study. Firstly, we do not know if the
317 original physical marker positions were optimal. Moving the markers
318 anteriorly/posteriorly may have in fact been moving them closer to the original target
319 positions. However, as the effect of systematic marker displacements on outcome
320 variables is unidirectional, the original marker locations will not affect our general
321 conclusions. Secondly, there is there is likely to be an element of random variation in
322 real-world marker placement, alongside the systematic element investigated here
323 (Osis et al. 2016). Random marker placement error and its effect on kinematics and
324 kinetics requires further research. Also, it is important to note that the specific errors
325 reported in this study are limited to the CoD task analysed, with marker placement
326 likely having a different effect in different tasks (Baker, Finney, and Orr 1999a).
327 Lastly, our marker displacements were simplistic in nature and do not directly mimic
328 real world marker placement error. We implemented fixed displacements, meaning
329 markers were moved the same distance relative to the original marker position
330 across all time points of the task. Physically moving markers across a range of ± 20
331 mm on the skin would involve a certain amount of medio-lateral in addition to
332 anterior/posterior displacement, as well as different soft tissue artefacts (STA).
333 Different STA's would alter the observed errors in this study, meaning translating our
334 findings directly to real world scenarios is challenging. Separating the effect of

335 marker placement error from that of STA is difficult and the relationship between
336 these two major sources of error is an area that warrants further research. For this
337 study, we chose to focus on simple anterior/posterior displacements, as the model
338 definitions indicate that these are the marker displacements that most substantially
339 effect model outputs (Kadaba, Ramakrishnan, and Wooten 1989). Accounting for the
340 additional effects of medio-lateral displacements and STA went beyond the scope of
341 the current investigation.

342 Alternative methods for modelling the human body have been developed to mitigate
343 the effect of STA and provide improved anatomical relevance compared to the CGM.
344 These include models that implement the calibration anatomical systems technique
345 (CAST), or models that allow for six degrees of freedom (6DOF) at each joint.

346 Models implementing CAST or 6DOF continue to work on the assumption that
347 marker placement is consistent and repeatable between practitioners (Charlton et al.
348 2004). Indeed, any model utilising anatomical markers to define joint centres and
349 segment orientations makes this assumption. At present no alternative model or
350 technique has been as widely implemented and validated as the CGM (Baker et al.
351 2017; Charlton et al. 2004). Research into the sensitivity of alternative modelling
352 techniques to marker placement, and how this compares to the CGM is required
353 prior to any widespread clinical application. While limited in certain aspects, the CGM
354 currently presents a practical, deterministic, extensively validated model that can be
355 easily implemented in routine clinical practice. These factors may explain the
356 continued widespread use of the CGM in contemporary biomechanical research
357 (Cortes, Onate, and van Lunen 2011; Gore et al. 2018; Lee, Chow, and Tillman
358 2014; B. Marshall et al. 2015; McLean, Huang, and Van Den Bogert 2005; Pollard,
359 Sigward, and Powers 2007a; Sigward and Powers 2007). When utilising the CGM

360 however, it should be done in a manner that openly acknowledges its limitations
361 within the context of the study aims and reported results. If attempting to identify
362 relatively small differences in frontal and transverse plane kinematics for example, it
363 should be made explicitly clear that any identified differences may be attributable to
364 instrumental error such as marker placement.

365 In conclusion, we have shown that systematic differences in the placement of the
366 THI, KNEE and TIB markers, within or bordering on reported inter-tester variability
367 ranges, can cause statistically significant differences in multiple kinematic and kinetic
368 variables across various periods of CoD stance. Many variables affected have
369 previously been associated with increased frontal plane knee loading during CoD,
370 which is considered a key risk factor for ACL injury. Errors were particularly
371 pronounced across the first 20% of stance, a period from which discrete kinematic
372 and kinetic variables are regularly reported. Our findings demonstrate the minimum
373 systematic differences in marker positions required to cause significant differences in
374 lower extremity kinematics and kinetics. These thresholds can be used by
375 laboratories to establish acceptable levels of inter-tester variability in marker
376 placement. If inter-tester variability is above these thresholds, statistical inferences
377 and corresponding clinical recommendations related to group differences should be
378 made with caution, as marker placement differences may result in invalid
379 conclusions.

380 **Conflict of interest statement**

381 The authors confirm that there is no financial or personal relationship with other
382 individuals or organisations that could inappropriately influence this work.

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