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Sex differences in ACL loading and strain during typical athletic movements: a musculoskeletal simulation analysis

Sinclair, Jonathan Kenneth, Brooks, Darrell and Stainton, Philip

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1 **Sex differences in ACL loading and strain during typical athletic movements: a**
2 **musculoskeletal simulation analysis.**

3 **Abstract**

4 **Purpose:** Female athletes experience anterior cruciate ligament (ACL) injuries at a much
5 greater rate than males, yet the mechanisms responsible for this are not well understood. The
6 current investigation aimed using a musculoskeletal simulation based approach, to examine
7 sex differences in ACL loading parameters during cut and hop movements.

8 **Methods:** Fifteen male and fifteen female participants completed 45° cut and maximal one
9 legged hop movements. Three-dimensional motion capture and ground reaction force data
10 during the stance phase of the cut movement and landing phase of the one legged hop were
11 obtained. Lower extremity muscle forces, ACL forces and ACL strains were extracted via a
12 simulation based approach using a musculoskeletal model, with an ACL insertion into the
13 femur and tibia.

14 **Results:** During the hop movement females were associated with significantly greater peak
15 ACL forces (male = 15.01 N/kg & female = 15.70 N/kg) and strains (male = 6.87 % &
16 female = 10.74 %). In addition, for both the cut (male = 4.45 & female = 1.45) and hop (male
17 = 2.04 & female = 1.46) movements the soleus/ gastrocnemius ratio was significantly larger
18 in males.

19 **Conclusions:** The current investigation provides new information regarding sex differences
20 during athletic movements that provide further insight regarding the increased incidence of
21 ACL injuries in females.

22
23 **Introduction**

24 Although engagement in regular physical activity and sport is associated with a variety of
25 physiological and psychological benefits (Warburton et al., 2006), participation in athletic
26 activity is allied to a high risk from musculoskeletal injury (Finch et al., 2001). The knee is
27 the most commonly injured musculoskeletal site (John et al., 2016), and the anterior cruciate
28 ligament (ACL) is the most frequently disrupted knee ligament (Evans et al., 2014). The ACL
29 is essential for the provision and maintenance of knee stability during dynamic activities
30 (Ellison et al., 1985). With its functional properties and complex anatomy, the ACL is acutely
31 competent in limiting both excessive anterior tibial translation and coronal/ transverse plane
32 knee movements (Dargel et al., 2007).

33

34 ACL injuries are predominantly, non-contact in nature, in that the structural integrity of the
35 ligament becomes compromised without physical contact between athletes (Boden et al.,
36 2010). Mechanically, ACL injuries occur when the ligament experiences excessive tensile
37 forces and strains (Smith et al., 2012). Aetiological analyses have shown that the ACL is
38 most vulnerable in the period following foot contact with the ground, in tasks involving
39 sudden decelerations, landings and cutting manoeuvres (Olsen et al., 2004). Athletes with
40 ACL rupture typically undergo reconstructive intervention using auto/ allografts to stabilize
41 the knee (Gottlob et al., 1999; Kaeding et al., 2015). Although the accelerated rehabilitation
42 program developed by Shelbourne et al., (1992) has significantly shortened recovery time
43 following surgery, ACL reconstruction is still preceded by a significant and aggressive period
44 of rehabilitation, with total allocated costs exceeding \$3.4 billion (Gottlob et al., 1999).
45 Importantly, the ACL is associated with poor healing capacity and the risk of a second injury
46 is as high as 30% in the ipsilateral knee (Di Stasi et al., 2013). ACL injuries frequently lead
47 to chronic knee pain, and athletes who experience an ACL pathology are as many as 10 times
48 more susceptible to early-onset degenerative knee osteoarthritis in (Øiestad et al., 2009),

49 leading not only to a decline in athletic participation but also enduring disability in later life
50 (Ajuied et al., 2014). Radiographic knee osteoarthritis significantly reduces health-related
51 quality of life, and degenerative joint disease secondary to ACL injury imposes a significant
52 economic burden (Mather et al., 2013).

53

54 Importantly, epidemiologic analyses have shown that female athletes have a 2-8 fold
55 increased risk of ACL pathology in relation to age-matched males of similar athletic ability
56 (Arendt et al., 1999). Increased ACL injury risk allied to enhanced participation in athletic
57 activities in females has fuelled a range of comparative and interventional biomechanical
58 investigations aimed at identifying modifiable risk factors. However, the precise aetiology of
59 ACL injury is currently disputed within clinical/ biomechanical literature, with some
60 advocating a predominantly sagittal plane ACL injury mechanism (Yu & Garrett, 2007), and
61 others supporting the notion that lower extremity coronal and transverse plane loads and
62 movements are also associated with ACL injury risk (Wascher et al., 1993; Markolf et al.,
63 1995; Krosshaug et al., 2007; Boden et al., 2009). Females have been proposed to exhibit
64 riskier landing mechanics during dynamic activities that are linked with ACL injury
65 (Voskanian, 2013). Indeed, three-dimensional kinetic and kinematic analyses have shown
66 that females exhibit reduced hip, knee and ankle flexion angles, enhanced knee valgus angles,
67 larger ground reaction forces (GRF), greater tibia anterior shear forces, larger knee extension
68 and valgus moments, greater hip internal rotation, hip adduction and knee rotation during
69 deceleration or landing manoeuvres (Decker et al., 2003; Malinzak et al., 2001; Chappell et
70 al., 2002; Lephart et al., 2002; Ford et al., 2003; Lin et al., 2012; Sinclair et al., 2012).

71

72 During single limb landing and deceleration activities, anterior tibial translation is primarily
73 restrained by the ACL, therefore the knee joint must be stabilized and protected from
74 excessive loads on the joint's soft tissue and ligaments (Quatman & Hewett, 2009). Muscle
75 recruitment patterns play a key role, and appropriate muscular preference, recruitment and
76 timing, are essential for the maintenance of knee joint stability (Li et al., 1999). As they span
77 the knee joint, the hamstring and quadriceps muscle groups are considered crucial in
78 moderating ACL loading (Shimokochi & Shultz, 2008). Indeed, numerous analyses have
79 revealed that the quadriceps serve to produce anterior tibial translation and thus increase ACL
80 loading, whereas the hamstring muscle group are act to oppose tibial translation and thus
81 attenuate ACL loads (Baratta et al., 1988; Solomonow et al., 1987; Draganich & Vahey,
82 1990; Durselen et al., 1995; Li et al., 1999; Markolf et al., 2004). Importantly, previous
83 analyses have shown that females exhibit quadriceps dominance during landing, and take
84 longer to generate maximum hamstring torque than their male counterparts (Hewett et al.,
85 1996; Huston et al., 1996). Several electromyographical analyses have confirmed this notion
86 using the hamstring/ quadriceps ratio. Females are habitually associated with lower values
87 than males, indicating greater relative involvement of the quadriceps in relation to the
88 hamstrings (Ebben et al., 2010; Landry et al., 2007; Nagano et al., 2007). This is also
89 considered a key mechanism that predisposes female athletes to ACL injury (Ruan et al.,
90 2017). In addition, recent analyses have also shown that muscles may not need to cross the
91 knee joint in order to contribute to ACL loading. Indeed, both Mokhtarzadeh et al., (2013)
92 and Adouni et al., (2016) have demonstrated the agonistic function of the soleus muscle in
93 ACL loading. However, there has yet to be any investigation to examine sex differences in
94 soleus muscle function during typical athletic movements.

96 Numerous prevention programmes have been devised in order to address mechanisms linked
97 to the aetiology of injury, which have had some success in attenuating the rate of ACL
98 injuries (Caraffa et al., 1996; Hewett et al., 1999; Myklebust et al., 2003; Mandelbaum et al.,
99 2005; LaBella et al., 2011). However, the efficacy of any intervention is dependent on a
100 sound comprehension of the underlying causative mechanisms of the associated condition,
101 and the aetiology for this gender discrepancy is not completely understood (Dai et al., 2014).
102 To date there has yet to be any investigation, which has examined sex differences in ACL
103 loading and strain parameters during athletic movements, principally due to the inability to
104 non-invasively quantify ACL loads and strain during high-risk athletic movements (Kar &
105 Quesada, 2012). Furthermore, there has also yet to be any investigation which has
106 concurrently examined sex differences in GRF's, three-dimensional knee kinematics and
107 muscle forces during athletic movements. However, advances in musculoskeletal simulation
108 software and enhancements in algorithmic complexity have led to the development of a
109 bespoke model with a six degrees of freedom at the knee joint and the inclusion of a passive
110 ACL inserted into the femur and tibial segments (Kar & Quesada, 2012). To date however,
111 this more advanced model has not yet been utilized to explore sex differences in ACL loading
112 and strain during high-risk athletic movements.

113

114 The aim of the current investigation was to examine sex differences in ACL loading, GRF's,
115 three-dimensional knee kinematics and muscle forces during cut and hop movements using a
116 musculoskeletal simulation based approach. In light of the increased incidence of ACL
117 pathologies in female athletes, the high likelihood of re-injury and the chronic reductions in
118 both musculoskeletal health and athletic functionality, it can be concluded that further insight
119 into the biomechanical differences between males and female athletes would be of both

120 practical and clinical significance. The current investigation tests the hypothesis that females
121 will be associated with greater ACL loading parameters during both cut and hop movements.

122

123 **Methods**

124 *Participants*

125 Fifteen male (age 30.1 ± 5.2 years, height 1.75 ± 0.1 m and body mass 77.1 ± 10.8 kg) and
126 fifteen female (age 29.6 ± 5.6 years, height 1.66 ± 0.1 m and body mass 65.8 ± 9.9 kg)
127 recreational athletes volunteered to take part in the current investigation. All participants
128 were free from lower extremity musculoskeletal pathology at the time of data collection and
129 had not undergone surgical intervention of the knee joint. All provided written informed
130 consent and ethical approval was obtained from the University of Central Lancashire, in
131 accordance with the principles documented in the declaration of Helsinki.

132

133 *Procedure*

134 Participants completed five repeats of two sport specific movements; one legged hop and 45°
135 cut. To control for any order effects the order in which participants performed in each
136 movement condition were counterbalanced. Kinematic information was obtained using an
137 eight camera motion capture system (Qualisys Medical AB, Goteburg, Sweden) using a
138 capture frequency of 250 Hz. To measure kinetic information an embedded piezoelectric
139 force platform (Kistler National Instruments, Model 9281CA) operating at 1000 Hz was
140 utilized. The kinetic and kinematic information were synchronously obtained and interfaced
141 using Qualisys track manager.

142

143 To define the anatomical frames of the thorax, pelvis, thighs, shanks and feet retroreflective
144 markers were placed at the C7, T12 and xiphoid process landmarks and also positioned
145 bilaterally onto the acromion process, iliac crest, anterior superior iliac spine (ASIS),
146 posterior superior iliac spine (PSIS), medial and lateral malleoli, medial and lateral femoral
147 epicondyles, greater trochanter, calcaneus, first metatarsal and fifth metatarsal. Carbon-fibre
148 tracking clusters comprising of four non-linear retroreflective markers were positioned onto
149 the thigh and shank segments. In addition to these the foot segments were tracked via the
150 calcaneus, first metatarsal and fifth metatarsal, the pelvic segment was tracked using the PSIS
151 and ASIS markers and the thorax segment was tracked using the T12, C7 and xiphoid
152 markers. Static calibration trials were obtained with the participant in the anatomical position
153 in order for the positions of the anatomical markers to be referenced in relation to the tracking
154 clusters/markers, following which those not required for dynamic data were removed.

155

156 Data were collected during the cut and hop movements according to below procedures:

157

158 *Cut*

159 Participants completed 45° sideways cut movements using an approach velocity of 4.0 m.s⁻¹
160 ±5% striking the force platform with their right (dominant) limb. Cut angles were measured
161 from the centre of the force plate and the corresponding line of movement was delineated
162 using masking tape so that it was clearly evident to participants. The stance phase of the cut-
163 movement was defined as the duration over > 20 N of vertical force was applied to the force
164 platform.

165

166 *Hop*

167 Participants began standing by on their dominant limb; they were then requested to hop
168 forward maximally, landing on the force platform with same leg without losing balance. The
169 arms were held across the chest to remove arm-swing contribution. The hop movement was
170 defined as the duration from foot contact (defined as > 20 N of vertical force applied to the
171 force platform) to maximum knee flexion. The hop distance for each participant was
172 established during practice trials, and the starting position was marked using masking tape.
173 Hop distance for each participant was extracted as the horizontal displacement of the foot
174 centre of mass from the initial position to the point of foot contact.

175

176 *Processing*

177 Dynamic trials were digitized using Qualisys Track Manager in order to identify anatomical
178 and tracking markers then exported as C3D files to Visual 3D (C-Motion, Germantown, MD,
179 USA). Data during the appropriate phases of each movement were exported from Visual 3D
180 into OpenSim 3.3 software (Simtk.org) using a custom pipeline that allowed the inverse
181 kinematics to be exported in order to match the degrees of freedom associated with the
182 experimental model in OpenSim. A previously utilized musculoskeletal model with 54
183 muscle-tendon units in 12 segments was adopted (Kar & Quesada, 2012). This model differs
184 from the traditional gait2354 approach in that a 6 degrees of freedom knee joint was included
185 alongside ACL ligament bundles which were modelled as non-linearly elastic passive soft
186 tissues with their proximal and distal ends inserted into the femur and tibia.

187

188 Firstly, using data from anatomical landmarks collected during the static calibration trials, the
189 model was scaled for each participant within OpenSim (Lerner et al., 2015). In accordance
190 with Kar & Quesada, (2012), muscle, tendon and ligament dimensions were scaled in the

191 same manner as body segments, from the static trial marker positions. Following this, we
192 performed a residual reduction algorithm (RRA) within OpenSim to reduce the residual
193 forces and moments in accordance with the recommendations of Lund & Hicks, (2013).
194 Following the RRA, the computed muscle control (CMC) procedure was then employed to
195 estimate a set of muscle force patterns allowing the model to replicate the required
196 kinematics.

197

198 From the above processing, the peak ACL force during the phases of each movement was
199 extracted and normalized by dividing the net values by body mass (N/kg) (Kar & Quesada,
200 2012). Further to this, the time taken from the instance of footstrike to peak ACL force (ms)
201 was also extracted for statistical analysis. In addition, the maximum ACL strain (%) was
202 calculated by dividing the maximum ligament bundle length during the dynamic trials by the
203 resting length, which was obtained during the static calibration trials (Kar & Quesada, 2012;
204 Taylor et al., 2013). Finally, forces of the rectus femoris, vastus intermedius, biceps femoris
205 long head (LH), biceps femoris short head (SH), gastrocnemius, sartorius, gracillis, tensor
206 fascia lata (TFL), tibialis anterior, tibialis posterior and soleus muscle groups were quantified
207 at the instance of peak ACL force following normalization to body mass (N/kg).

208

209 **Quadriceps dominance in relation to the hamstring has been shown through**
210 **electromyographical analyses to be prominent in females (Ebben et al., 2010; Landry et al.,**
211 **2007; Nagano et al., 2007) and identified as a risk factor for ACL injury (Ruan et al., 2017).**
212 **Musculoskeletal simulation analyses are able to generate outputs for individual knee extensor**
213 **and flexor muscles (Delp et al., 2007). Therefore, they have the potential to offer further**
214 **insight regarding sex differences in specific extensor and flexor muscle-tendon unit outputs,**

215 which may provide more detailed information regarding the role of muscular dominance in
216 females. As such, flexor (biceps femoris LH, biceps femoris SH, gastrocnemius, Sartorius
217 and gracillis) and extensor (rectus femoris and vastus intermedius) ratios were also calculated
218 at the instance of peak ACL force. Finally, as the soleus muscle has been proposed as a
219 mechanism by which the ACL is protected during landing manoeuvres in relation to the
220 gastrocnemius (Mokhtarzadeh et al., 2013), the soleus/ gastrocnemius ratio was also
221 quantified at the instance of peak ACL force.

222

223 In addition to the aforementioned muscle analyses, three dimensional knee joint angular
224 kinematic measures were also examined. Knee joint kinematic parameters that were extracted
225 for statistical analysis were 1) angle at foot contact, 2) peak angle and 3) angular range of
226 motion (ROM) from foot contact to peak angle. Furthermore, the hip flexion angle at the
227 instance of foot contact was also extracted for further analysis. Finally, vertical and anterior-
228 posterior GRF's were quantified at the instance of peak ACL force following normalization
229 to body mass (N/kg).

230

231 *Analyses*

232 Descriptive statistics of means and standard deviations (SD) were obtained for each outcome
233 measurement. Shapiro-Wilk tests were used to screen the data for normality. For the cut
234 movement, sex differences in ACL loading and muscle force parameters were examined
235 using univariate ANOVA's. In addition, as hop distance was statistically larger in male
236 athletes (1.66 ± 0.11 m) compared to females (1.32 ± 0.17 m), sex differences in ACL and
237 muscle forces were examined using a univariate ANCOVA with hop distance as the
238 covariate. This was undertaken due to the increased vertical and anterior-posterior GRF's

239 associated with greater landing distances (Barker et al., 2017). Statistical significance
240 throughout was accepted at the $P \leq 0.05$ level, and effect sizes were calculated using partial
241 Eta^2 ($\rho\eta^2$). All statistical actions were conducted using SPSS v24.0 (SPSS Inc, Chicago,
242 USA).

243

244 **Results**

245 *Cut movement*

246 The soleus/ gastrocnemius ratio at the instance of peak ACL force was significantly larger in
247 males (Table 1). In addition, knee peak valgus, internal rotation and internal rotation ROM
248 were shown to be significantly larger in females (Table 2).

249

250 **@@@TABLE 1 NERE HERE@@@**

251 **@@@TABLE 2 NERE HERE@@@**

252

253 *Hop movement*

254 For the hop movement, females were associated with significantly increased peak ACL
255 forces and peak ACL strains (Table 3). In addition, the soleus/ gastrocnemius ratio at the
256 instance of peak ACL force was significantly larger in males (Table 3). Finally, knee peak
257 valgus and internal rotation were shown to be significantly larger in females (Table 4).

258

259 **@@@TABLE 3 NERE HERE@@@**

260 **@@@TABLE 4 NERE HERE@@@**

261

262 **Discussion**

263 The aim of the current investigation was to examine sex differences in ACL loading
264 parameters during cut and hop movements. To the authors' knowledge, this represents the
265 first investigation to quantify ACL forces and strains in male and female athletes using a
266 musculoskeletal simulation based approach. Given the debilitating nature of ACL
267 pathologies, the high incidence of re-injury and the increased susceptibility to degenerative
268 joint disease secondary to ACL injury, a study of this nature may provide important
269 information to inform future prevention and rehabilitation programmes.

270

271 For the cut movement, the current investigation provided scant support for the hypothesis in
272 that although very small increases in ACL loading parameters were noted in female athletes,
273 the differences did not reach statistical significance. For the more dynamically and
274 functionally challenging hop movement however, the findings support our hypotheses as both
275 peak ACL force and ACL strain were shown to be statistically larger in females when
276 adjusted for the influence of hop length through covariate analyses. This concurs with the
277 observations of Schilaty et al., (2018), who showed using cadaveric impacts that female
278 ligaments experience greater strain than males during a simulated landing task. Mechanically,
279 ACL injuries occur when the ligament experiences excessive tensile forces and strains.
280 Therefore, given the statistical differences between sexes during the hop movement and with
281 the ACL strain being larger in female athletes, this finding may provide biomechanical
282 insight regarding the aetiology of injury in females.

283

284 Female athletes are believed to exhibit riskier biomechanics and increased quadriceps
285 dominance during landing (Voskanian, 2013). The kinematic observations from the current
286 investigation support the aforementioned notion, as females were associated with statistically
287 greater coronal and transverse plane knee joint kinematics during both movements. Increases
288 in knee valgus have been reported previously (Ford et al., 2003; Russell et al., 2006;
289 Kernozek et al., 2005), and may be pertinent in relation to the increased incidence of ACL
290 injury in females. Prospective analyses show that athletes experiencing ACL injury exhibited
291 knee valgus angles $\geq 8^\circ$ than those who remained uninjured (Hewett et al., 2005).
292 Furthermore, following ACL rupture, lateral epicondyle bone bruises are evident in 80% of
293 cases, further implicating the valgus position of knee joint in relation to the aetiology of ACL
294 pathologies (Viskontas et al, 2008). In addition, increased knee internal rotation in female
295 athletes agrees with previous analyses (Kiriyaama et al., 2008; Sinclair et al., 2012), and given
296 recent observations may be clinically meaningful regarding the increased likelihood of
297 ACL injuries in females. Based on video analyses of ACL ruptures post injury, it was initially
298 proposed that external rotation was the transverse plane knee mechanism responsible for
299 ACL injuries (Ebstrup & Bojsen-Molle, 2000). However, Koga et al., (2010) and Koga et al.,
300 (2011) have shown that the knee exhibits internal rotation until ligament failure, following
301 which the direction of knee rotation reverses. Therefore, prophylactically attenuating knee
302 valgus and internal rotation measures in female athletes either using movement re-training or
303 via external supports should remain a key objective for trainers and physical therapy
304 professionals alike.

305

306 Furthermore, in addition to riskier biomechanics females are purported to exhibit increased
307 quadriceps dominance during landing (Voskanian, 2013). Previous electromyographical
308 analyses have revealed a diminished hamstring/ quadriceps ratio in females (Nagano et al.,

309 2007). The current investigation is the first to explore potential quadriceps dominance in
310 females using muscle forces provided by musculoskeletal simulation. However, the findings
311 from the current study do not appear to support the aforementioned concept of quadriceps
312 dominance in female athletes. Firstly, there were no statistical sex differences in quadriceps
313 muscle forces and secondly none of the sex differences in any of the quadriceps muscle force
314 ratio's reached statistical significance.

315

316 Importantly, the musculoskeletal model utilized in this investigation also quantified both
317 soleus and gastrocnemius forces. The kinetics of these two muscles are typically ignored in
318 analyses concerning the loads experienced by the ACL owing of the supposition that they
319 have limited influence due to the muscles lines of action being close to the long axis of the
320 tibia (Mokhtarzadeh et al., 2013). However, previous modelling analyses by Mokhtarzadeh et
321 al., (2013) and Adouni et al., (2016) have shown that the soleus protects the ACL during
322 landing manoeuvres by exerting a posterior force on the tibia and that the gastrocnemius acts
323 as an ACL antagonist. The current investigation showed that the muscle force ratio between
324 the soleus and gastrocnemius muscles was statistically larger in male athletes, indicating a
325 more favourable ratio in terms of protection from ACL injuries during high intensity athletic
326 movements.

327

328 A potential limitation to the current investigation is the mechanism by which the simulation
329 analyses were conducted. Although a powerful tool that has been utilized in previous
330 analyses to simulate ACL mechanics (Kar & Quesada, 2013), the CMC processes is
331 insensitive to variations in muscle activation and limited in its ability to quantify muscle
332 coordination during dynamic tasks (Zajac et al., 2002). As both of these parameters have

333 been shown previously to exhibit both sex and movement differences (Nagano et al., 2007),
334 this may represent a methodological drawback to the current study. In addition, the lack of
335 sex specificity in regards to the anatomy and scaling of the ACL may serve as a limitation to
336 this investigation. As the ACL contributes significantly to knee joint load bearing and
337 stability, incorporation of a sex specific scaling mechanism may improve the efficacy of
338 musculoskeletal simulation analyses concerning the knee joint. That ACL strain was
339 quantified by standardizing ligament elongation to a resting length obtained during the static
340 calibration trial, may also represent a drawback to this instigation. Although this procedure
341 was selected in accordance with Kar & Quesada, (2012) and Taylor et al., (2013), due to the
342 complications associated with determining an accurate in vivo resting length (Fleming and
343 Beynnon, 2004) and there remains some uncertainty regarding the accuracy of true strain
344 values. Finally, as three-dimensional knee kinematics were quantified using skin mounted
345 markers this may serve as a limitation. Particularly in light of the findings provided by Benoit
346 et al., (2006) indicating that kinematic waveforms produced using this technique may not be
347 representative of the motion of the underlying bones.

348

349 In conclusion, the current investigation adds to the current literature by exploring sex
350 differences in ACL loading, GRF's, three-dimensional knee kinematics and muscle forces
351 using a musculoskeletal simulation based approach. Importantly, the findings from this study
352 showed that during the hop movement, females were associated with significantly greater
353 peak ACL forces and strains. In addition, for both movements the soleus/ gastrocnemius ratio
354 at the instance of peak ACL force was significantly larger in male athletes. Therefore, the
355 current investigation provides new information regarding sex differences during athletic
356 movements that provide further insight regarding the increased incidence of ACL injuries in
357 females.

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