

1 **Abstract**

2 Objectives

3 To investigate the influence of BMX helmets and neck braces on translational and
4 rotational accelerations in youth riders.

5 Design

6 Mixed model, repeated measure and correlation.

7 Methods

8 Twenty three competitive youth BMX riders classified by age group (6-9 yrs, 10-13 yrs
9 and 14-18 yrs) completed 6 laps of an indoor BMX track at race pace, 3 laps without a
10 neck brace (NB) and 3 without brace (WB). A triaxial accelerometer with gyroscope was
11 placed behind the right ear to determine the mean number of accelerations, translational
12 and rotational, of the head between conditions and by age group.

13 Results

14 Significant reductions by condition ($p = 0.02$) and by age ($p = 0.04$) were found for the
15 number of accelerations, though no interactions (condition \times age) were revealed.
16 Significant increases by age ($p = 0.01$) were revealed for translational accelerations, whilst
17 significant increases by condition ($p = 0.02$) were found for rotational accelerations. In
18 addition, significant correlations were revealed between relative helmet mass and age (r
19 $= 0.83$; $p 0.001$) and relative helmet mass and number of accelerations ($r = 0.46$; $p = 0.03$).

20 Conclusions

21 Accelerations at the head decreased with increased age, possibly due to the influence of
22 greater stabilising musculature. Additionally, neck braces also significantly reduced the
23 number of accelerations. However, the magnitude of accelerations may be influenced by
24 riding dynamics. Therefore, the use of neck braces combined with strength work to

25 develop neck strength, could aid in the reduction of head accelerations in youth BMX
26 riders.

27

28 Keywords: Injury; accelerometry; concussion; cycling.

29

30 **1. Introduction**

31 Bicycle motocross (BMX) has been an Olympic sport since 2008 and involves up
32 to eight riders competing against each other in qualifying heats¹. Courses are typically
33 between 200 m and 400 m in length and require riders to negotiate a variety of straight
34 flat sections, jumps and banked corners. Races generally last between 30 and 50 s and
35 demand a high anaerobic endurance capacity^{2,3}.

36 Though not considered a contact sport, the high speeds, close proximity of riders
37 and large jumps present considerable potential for injury. Few published studies exist on
38 the prevalence of injury, and the types of injuries sustained, during BMX riding.
39 Engebretsen et al. (2013)⁴ reported that during the 2012 Olympic Games, all 48 of the
40 registered BMX riders sustained an injury of some form during training or competition.
41 Though they didn't state the exact number or percentage break down of injuries for BMX
42 specifically, Engebretsen et al. stated that the majority of injuries across all sports were
43 musculoskeletal in nature, yet one incident of concussion was also reported for BMX.
44 Additionally, 25 % of reported injuries across all Olympic sports were attributed to overuse
45 injuries, 20 % were due to non-contact trauma and 14 % due to contact with other athletes,
46 though again the specific breakdown for BMX was not stated. The potential for head
47 injuries, notably concussions and mild traumatic brain injuries, may be elevated for BMX
48 given the nature of this event. To date, there is little information available on head injuries
49 sustained during BMX riding, nor attempts to profile the biomechanics of head movements
50 during training and competition.

51 BMX riders are required by the world governing for cycling (Union Cycliste
52 Internationale) body regulations to wear full-face motocross style helmets. Such helmets
53 have previously been shown to significantly reduce the frequency and severity of head
54 and brain injuries resulting from bicycle crashes^{5,6,7}. However, these helmets are generally
55 much heavier than normal open face bicycle helmets (~300 g), with a mass typically
56 between 900 g and 1700 g based on manufacturer claims⁸. Though the additional mass
57 may not be an issue for adult riders, who are more physically mature, it may result in
58 additional neck loading in younger, less developed riders as a result of increased helmet
59 mass relative to body mass. Greater neck strength, allied to activating the neck muscles
60 in readiness for impact, have been proposed to reduce an athlete's risk of concussion
61 during a collision^{9,10}. Riders with smaller and weaker necks are suggested to be more
62 likely to experience greater translational and rotational displacements of the head
63 following impact¹¹. However, this relationship remains somewhat inconclusive.

64 In addition to the helmet, riders can also wear a protective neck brace, though this
65 is not mandatory. These devices were designed to reduce translational and inclination
66 accelerations of the head, by transferring the accelerations from the head and neck to the
67 torso, but without compromising rotational range of movement (ROM)¹². However, in
68 motocross riding, Thiele et al. (2016)¹³ showed neck braces reduced activity in the primary
69 neck muscles, along with a reduction in range of motion, both translational and rotational.
70 To date though, no attempt has been made to review acceleration of the head when using
71 such braces during BMX riding.

72 Therefore, the aims of this study were to identify the number of accelerations, the
73 magnitude of translational and rotational neck accelerations during BMX in different
74 chronological age groups; to determine the influence of wearing a neck brace on these
75 accelerations and to determine range of motion (ROM) with and without helmet and neck
76 brace. The study also aimed to determine whether any relationships existed between the
77 number of accelerations, magnitude of accelerations, rider age and helmet mass relative

78 to body mass (RHM). It was hypothesised that neck accelerations would be greatest in
79 younger riders and that the neck brace would reduce the magnitude of accelerations
80 without affecting neck ROM. Finally, given the heavier RHM, it was hypothesised that
81 relationships would exist between this and age, number of accelerations and the
82 magnitude of accelerations.

83

84 **2. Methods**

85 Twenty-three competitive BMX cyclist participated in the study. All had previous
86 experience of riding the track used for testing (National Cycling Centre indoor BMX Track,
87 Manchester, UK). Riders were placed into three groups based on chronological age, and
88 classified as 6-9 yrs (N=8; mean age 7.00 ± 1.07 yrs, body mass 28.33 ± 4.53 kg, stature
89 129.11 ± 6.77 cm); 10-13 yrs (N=8; mean age 11.88 ± 1.25 yrs, body mass 47.79 ± 8.26
90 kg, stature 153.36 ± 9.21 cm); 14-18 yrs (N=7; mean age 15.57 ± 1.72 yrs, body mass
91 61.10 ± 10.24 kg and stature 167.27 ± 6.88 cm). Written and informed consent was
92 obtained from the participants and parent/guardians prior to the study. The study was
93 granted ethical approval from the University of Derby Ethics Human Studies Board, and
94 was in accordance with the principles outlined in the Declaration of Helsinki.

95 The track was a national standard, indoor BMX track and had a 5 meter high start
96 ramp with a 28° decent angle. Track length was 400 meters and consisted of three banked
97 corners (berms) and four straight sections with a number of technical jumps on each
98 straight. Riders performed three laps of the track without a neck brace (NB) and three laps
99 with a neck brace (WB). The neck braces (Atlas, Atlas Brace Technologies, Valencia,
100 USA) came in three sizes based on chest size (53-63 cm, 61-71 cm and 74-84 cm) and
101 weighted 375 g, 460 g and 590 g, respectively. The manufacturers' guidelines for fitting of
102 the neck braces was followed. This first required measuring the chest circumference at
103 the level of the axilla and selecting the appropriate neck brace for that size. Secondly, the

104 rear positioning mounts on the neck brace were adjusted to ensure the chest, back and
105 shoulder pads sat flush against the body for each participant. As the neck brace was not
106 directly attached to the helmet and was fitted to minimise movement around the neck and
107 shoulder complex, it was deemed unlikely to contribute to translational or rotational
108 accelerations of the head. Participants helmets were also weighed using a digital scale
109 (Salter, Kent, UK) to the nearest 0.1 g, in order to determine RHM (g/Kg BM). A triaxial
110 accelerometer with gyroscope (xPatch, X2 Biosystems, Seattle, USA) was used to
111 measure the magnitude of translational (g) and rotational (rads/s^2) accelerations of the
112 neck along with the number of accelerations for each trial. Sensors were positioned behind
113 the right ear at the level of the occipito-temporal suture (Fig. 1). Separate sensors were
114 used for the NB and WB trials for each rider. Translational accelerations were sampled
115 at 1000 Hz, whilst rotational accelerations were sampled at 800 Hz. The minimum
116 recording threshold was set to 5 g, whilst the sensors had a refresh rate of 100 Hz. The
117 xPatch system had been validated previously for accelerations up to 160 g^{14} . Any values
118 recorded either above or below the minimum and maximum thresholds were deemed
119 erroneous or 'clack' accelerations by the proprietary software (X2 Biosystems Injury
120 Management Software) and removed from the dataset. As all riders were familiar with the
121 track already, a 10 min warm up period was given prior to starting data collection. Riders
122 were then instructed to ride full laps of the track as quickly as possible, without stopping,
123 before returning to the start gate for a 5 min passive recovery between laps. The order of
124 the trials were randomised and conducted over a three week period.

125 Range of motion (ROM) of the cervical spine, in all conditions, was assessed using
126 simple 2D image processing. No participant reported any neck or spine discomfort, nor
127 had any musculoskeletal impingement at the time of measurement. A high quality digital
128 camera (Nikon D5600) was mounted on a tripod, approximately 3 meters from the
129 participant. In a seated position, with the head held in a neutral position, each participant
130 performed three sequential flexion and extension movements. The head returned to the

131 neutral position between each repetition. For the determination of ROM, post-processing
132 of 2D images was undertaken using open-source software (ImageJ,
133 <https://imagej.nih.gov/ij/>). With a perpendicular rule used for reference, an approximate
134 line was plotted between the tragion and the orbitale. This was used to determine a
135 neutral, or initial, angle. Deviation from this neutral angle, in both flexion and extension,
136 was then calculated as the ROM, in all un-helmeted and helmeted conditions. ROM
137 rotation measurements were determined in the supine position, again using a
138 perpendicular rule for reference. Three sequential rotations to the right and left sides were
139 completed. An approximate line was plotted between the bregma and nasal ridge. Again,
140 deviation from the neutral angle, in both right and left rotation, was calculated as the ROM.
141 For ROM movements, the average of the three measurements was calculated. To ensure
142 objectivity, a second assessor, who was blinded to the previous measurements, also
143 determined angles in post-processing. The coefficient of variation for cervical flexion was
144 1.6 %, extension 1.4 %, right rotation 2.1 % and left rotation 1.3 %. To determine intra-
145 tester reliability, repeated measures were undertaken on two separate days. For all items,
146 intraclass coefficients were ≥ 0.85 . Assessment of lateral flexion was performed but not
147 reported. This was due to the younger riders often being unable to maintain lateral flexion
148 in the correct alignment when helmeted, potentially due to the increased mass of the
149 helmet. As such, further kinematic analysis of the influence of helmet mass on cervical
150 ROM should be advocated.

151 All data were analysed using the statistical software package SPSS (version 23
152 SPSS Inc., Chicago, IL). The alpha level was set at $p \leq 0.05$. Differences in accelerations
153 and the number of accelerations between age groups and neck brace condition (NB vs
154 WB) were determined using mixed model repeated measure ANOVA's (Condition x Age).
155 *Post-hoc* analysis of within-subject effects were determined using a Bonferroni correction.
156 Differences in RHM (g/kg BM) by age group were analysed using a one-way repeated
157 measures ANOVA. Effect sizes were calculated using a partial Eta² (η_p^2). Effects sizes

158 were identified as; small = 0.01, medium = 0.06 and large = 0.14¹⁵. Pearson's product
159 moment correlations were used to determine any relationships between variables. Data
160 are reported as mean \pm SD (95 % CI) over the three laps for each condition unless
161 otherwise stated.

162

163 ***Figure 1 near here***

164

165 **3. Results**

166 Table 1 outlines the range of motion of the cervical spine by age group. Significant
167 interactions (condition x age) were found for cervical flexion ($F(2,17) = 15.41$; $p = 0.002$;
168 $\eta_p^2 = 0.49$) and extension ($F(2,17) = 5.15$; $p = 0.003$; $\eta_p^2 = 0.51$). For cervical flexion, post-
169 hoc comparisons revealed differences by age between the 6-9 and 10-13 ($p = 0.005$), and
170 6-9 and 14-18 ($p = 0.003$) age groups. No significant differences were found between the
171 10-13 and 14-18 years of age groups for any ROM variable ($p > 0.05$). In extension,
172 significant differences were noted between the 6-9 and 14-18 ($p = 0.02$) age groups. No
173 further significant differences were noted between any ROM variable, by condition or age.

174

175 ***Table 1 near here***

176

177 Table 2 summarises the findings for RHM, number of accelerations, translational
178 and rotational accelerations for each age group and for the NB and WB conditions. A
179 significant difference, $F(2,23) = 26.76$; $p < 0.001$; $\eta_p^2 = 0.73$, was found for RHM by age
180 group. No significant interactions (condition x age) were found for the number of
181 accelerations, though there were significant main effects for condition ($F(1,20) = 6.00$; p
182 $= 0.02$; $\eta_p^2 = 0.23$) and for age ($F(2,20) = 3.51$; $p = 0.04$; $\eta_p^2 = 0.26$). However, when *post-*

183 *hoc* comparisons were performed they didn't reveal differences between individual age
184 groups.

185 No interaction effect or main effect by condition were identified for translational
186 accelerations. However, there was a significant main effect for age ($F(2,20) = 5.55$; $p =$
187 0.01 ; $\eta_p^2 = 0.36$). When *post-hoc* comparisons were performed, they revealed significant
188 differences between the 6-9 yrs and 14-18 yrs age groups ($p = 0.04$) and the 10-13 yrs
189 and 14-18 yrs age groups ($p = 0.02$). Similarly, no interaction effect was found for
190 rotational accelerations. However, unlike with translational accelerations, a significant
191 main effect was found for condition ($F(1,20) = 7.15$; $p = 0.02$; $\eta_p^2 = 0.26$), but not for age.

192 Significant relationships were found between RHM and age ($r = 0.83$; $p = 0.001$)
193 and RHM and the number of accelerations in the NB condition ($r = 0.46$; $p = 0.03$). No
194 other significant relationships were found.

195

196 ***Table 2 near here***

197

198 **4. Discussion**

199 The results of this study found that the number of accelerations observed at the
200 head, above the pre-determined threshold, were significantly reduced with the use of a
201 neck brace. In addition, there was a significant main effect for age, with the number of
202 accelerations decreasing with increasing age. This could be attributed to increased
203 muscular development about the neck and shoulders with age, to help dampen external
204 loading of stabilising neck musculature. It would be expected that neck flexor, extensor
205 and stabilising rotational musculature of the shoulders would accommodate such rapid
206 head movements. Though neck and shoulder muscularity were not directly determined in

207 this study, future studies might seek to evaluate these and their potential influence on
208 head accelerations.

209 Decreasing translational and rotational head accelerations has been proposed via
210 a number of key mechanisms, notably when related to heading of soccer balls^{16,17}. These
211 include better alignment of the head-neck-torso, increasing neck flexor and extensor
212 strength and enhancing neuromuscular control of the key stabilising muscles. It is likely
213 that such interventions may have utility in improving stabilisation and dampening
214 properties of the head, thereby reducing accelerations, yet this remains to be investigated.
215 The populations tested in the current study will have certainly encompassed prepubertal,
216 circumpubertal and late maturing individuals. In such young populations, generally, overall
217 strength has yet to develop. When adding in the confounding effect of additional mass to
218 the head, in the form of helmets, a clear potential for poor stabilisation of the head may
219 manifest. Strengthening and muscular recruitment activities to help stabilise the head may
220 be of value, and have been supported in literature elsewhere^{18,19}.

221 Though the magnitude of the transitional accelerations did not differ significantly
222 with or without the use of a neck brace, they were significantly different by age group, with
223 the eldest group eliciting the highest accelerations. It was observed that the younger riders
224 had a greater tendency to roll over the jumps with the wheels remaining in contact with
225 the ground, whilst those in the older group generally carried more speed into the jumps
226 and attempted to clear the jump by getting airborne. This in part, may have contributed to
227 the higher translational accelerations seen in the 14-18 yrs age group, because of greater
228 loading upon landing. However, further analysis is needed to quantify this. Additionally,
229 whilst beyond the scope of this study, it may be of interest for future studies to determine
230 the stiffness and magnitude of deformation of different neck braces to determine whether
231 this could influence the dampening of the accelerations and therefore the magnitudes of
232 the accelerations.

233 Despite this, it is of interest to note, that across all three age groups the
234 translational accelerations observed were much greater than those previously reported
235 for other sports. Lynall et al. (2016)²⁰ reported mean transitional loads of 12.51 g during
236 collegiate level women's soccer, with the mean number of accelerations per 90 minutes
237 of play ranging from 3.39 to 9.40, depending upon positional role. Participants in the
238 present study experienced translational accelerations between 20.4 and 29.6 g, whilst the
239 number of accelerations was more than double (6.4 to 17.5) those of the Lynall et al.
240 study, yet in less than a 50 s period. Similarly, research into head accelerations in
241 professional rugby league players also reported translational accelerations considerably
242 lower (~15 g) than in the present study²¹. These findings demonstrate the scale of head
243 accelerations during BMX riding, and in particular the severity of the accelerations with
244 which young riders are exposed to. This may have implications for potential brain injuries
245 and function. McAllister et al. (2014)²², measured cognitive function along with using
246 diffusion weighted imaging to determine brain white matter integrity and found both
247 cognition and white matter integrity were impaired with repetitive impacts as low as 33.4g
248 in soccer and ice hockey players. These impacts are comparable to those reported in the
249 present study. Therefore, any means to reduce these accelerations should be welcomed
250 by riders and governing bodies.

251 With respect to rotational accelerations, age did not significantly influence the
252 magnitude of the accelerations. There was a significant difference between the NB and
253 WB conditions though. However, whilst the use of a neck brace was shown to reduce the
254 number of translational accelerations, the opposite was observed for rotational values.
255 This is in opposition to the hypotheses. Though it is difficult to identify why the use of a
256 neck brace would increase rotational accelerations, one possible explanation may relate
257 to the riders perception of wearing them. Anecdotal conversations with the riders revealed
258 the majority of them stated they felt it restricted their head movement. As such, it may be
259 possible the riders overcompensated for the perceived limitation by consciously turning

260 the head more when wearing the brace. This may have resulted in the higher rotational
261 accelerations observed. However, further analysis is warranted to confirm this and to
262 determine whether a learning effect would influence the results with greater practice with
263 the neck brace. Our simple assessment of cervical range of motion revealed that a
264 relatively consistent increase in range of motion, both translational and rotational,
265 accompanied the wearing of a helmet, across all age groups. Peculiarly, for the very
266 youngest group cervical flexion decreased when a helmet was worn. This is likely to be
267 associated with the design of the full-face helmet, notably the pronounced chin area. This
268 may have created a restriction on full range of motion when contacting with the upper
269 sternoclavicular area. Introducing a neck brace did reduce range of motion in all cervical
270 movements. This was expected, yet not significant.

271 Correlative analysis revealed significant relationships between RHM and age and
272 between RHM and the number of accelerations in the NB condition, with a greater number
273 of accelerations observed in the youngest group. This again suggests that as riders age
274 and develop greater neck and shoulder musculature, this may aid in resisting neck
275 accelerations as a result of helmet mass. No further relationships were found either in the
276 NB or WB conditions. Once again this would seem to suggest that the use of a neck brace
277 could effectively negate the negative effects of increased helmet mass relative to body
278 mass.

279

280 **5. Conclusions**

281 This study found that BMX riders are exposed to high head accelerations
282 regardless of age group when compared to other sports. Our findings show that the
283 number of accelerations decreased with age, possibly as a result of muscular
284 development about the neck and shoulders. It would also appear that the use of a neck
285 brace could effectively further reduce the number of head accelerations across all age

286 groups. However, the magnitude of these accelerations may be more related to riding
287 dynamics and negative pre-conceptions relating to the wearing of neck braces. Lastly,
288 RHM also appears to be influential in the number of accelerations observed. Therefore,
289 the use of BMX helmets may place additional stress on the head and neck of younger
290 riders.

291

292 **Practical implications**

- 293 • Development of neck/shoulder strength might help reduce the number of
294 accelerations when not wearing a neck brace in younger riders.
- 295 • Neck braces can be used to effectively reduce the number of accelerations at the
296 head.
- 297 • Further familiarisation with the wearing of neck braces may be required to reduce the
298 possibility of over exaggerating rotational movement and therefore accelerations of
299 the neck.

300

301 **Conflict of interest**

302 The authors have no conflicts of interest related to this paper.

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375 **Table 1. Mean ± SD (CI) cervical range of motion, in un-helmeted, helmeted and**
 376 **helmet/brace conditions, by age group.**

	6-9 yrs	10-13 yrs	14-18 yrs
Un-helmeted Flexion (deg)	114.5 ± 9.9 (104.1-124.9)	135.9 ± 11.4 (126.4-145.5)	146.3 ± 8.7 (132.4-160.2)
Un-helmeted Extension (deg)	81.4 ± 19.7 (60.7-102.1)	58.7 ± 6.7 (53.1-64.5)	38.5 ± 2.9 (29.3-47.8)
Un-helmeted Right Rotation (deg)	62.7 ± 15.8 (46.1-79.3)	56.1 ± 11.5 (46.4-65.6)	41.6 ± 8.6 (29.1-55.2)
Un-helmeted Left Rotation (deg)	72.9 ± 13.2 (59.1-86.8)	56.5 ± 10.6 (47.5-65.4)	43.6 ± 9.1 (29.1-58.1)
Helmeted Flexion (deg)	108.1 ± 9.6 (98.8-119.1)	142.6 ± 11.7 (132.7-152.4)	158.5 ± 7.1 (147.3-169.8)
Helmeted Extension (deg)	82.9 ± 11.2 (61.7-104.1)	61.9 ± 6.4 (56.6-67.3)	42.7 ± 10.6 (25.7-59.6)
Helmeted Right Rotation (deg)	67.5 ± 11.1 (53.6-81.3)	65.4 ± 13.5 (49.9-80.8)	57.6 ± 16.9 (40.7-84.5)
Helmeted Left Rotation (deg)	73.7 ± 11.8 (61.3-86.2)	67.0 ± 21.2 (49.3-84.7)	57.8 ± 13.1 (58.7-75.7)
Brace and Helmeted Flexion (deg)	132.1 ± 4.6 (127.1-136.9)	132.4 ± 10.6 (123.4-141.4)	144.1 ± 13.5 (119.4-168.7)
Brace and Helmeted Extension (deg)	42.7 ± 6.4 (36.0-49.4)	54.3 ± 9.1 (46.8-62.0)	54.6 ± 10.5 (37.9-71.2)
Brace and Helmeted Right Rotation (deg)	48.6 ± 13.3 (24.1-73.1)	52.3 ± 12.5 (34.1-72.1)	50.1 ± 10.6 (21.1-80.3)
Brace and Helmeted Left Rotation (deg)	47.6 ± 10.8 (36.2-58.9)	51.1 ± 17.4 (28.2-74.1)	46.1 ± 14.4 (23.1-69.1)

377 **Table 2. Mean \pm SD (CI) number of accelerations, translational and rotational**
 378 **accelerations by age group over three laps.**

	6-9 yrs	10-13 yrs	14-18 yrs
Relative helmet mass (RHM) (g/kg)	40.1 \pm 5.3 (35.7-44.6)	25.6 \pm 7.1 (19.6-31.5)	19.4 \pm 4.0 (15.7-23.1)
<i>No neck Brace (NB)</i>			
Number of accelerations	17.5 \pm 7.3 (11.4-23.6)	13.9 \pm 2.9 (11.5-16.2)	12.7 \pm 5.0 (8.1-17.4)
Translational acceleration (g)	23.2 \pm 4.2 (19.6-26.7)	23.3 \pm 5.1 (19.1-28.1)	29.6 \pm 4.1 (25.7-33.3)
Rotational acceleration (rads/s ²)	1919.8 \pm 496.3 (1504-2334)	1440.7 \pm 471.2 (1047-1835)	1951.8 \pm 718.1 (1287-2616)
<i>With neck Brace (WB)</i>			
Number of accelerations	14.9 \pm 11.8 (5.1-24.8)	6.4 \pm 3.25 (4.1-9.1)	9.3 \pm 4.3 (5.3-13.2)
Translational acceleration (g)	22.3 \pm 7.7 (16.1-28.7)	20.4 \pm 8.3 (13.4-27.3)	28.9 \pm 9.1 (20.5-37.3)
Rotational acceleration (rads/s ²)	2769.2 \pm 1601.5 (1430-4108)	3178.1 \pm 1387.8 (2018- 4338)	1988.4 \pm 935.6 (1123-2854)

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