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Profiling of translational and rotational head accelerations in youth BMX with and without neck brace

Abstract

Objectives

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

Design

Mixed model, repeated measure and correlation.

Methods

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

Results

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

Conclusions

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

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

Keywords: Injury; accelerometry; concussion; cycling.

1. Introduction

Bicycle motocross (BMX) has been an Olympic sport since 2008 and involves up to eight riders competing against each other in qualifying heats[1]. Courses are typically between 200 m and 400 m in length and require riders to negotiate a variety of straight flat sections, jumps and banked corners. Races generally last between 30 and 50 s and demand a high anaerobic endurance capacity[2,3].

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

BMX riders are required by the world governing for cycling (Union Cycliste Internationale) body regulations to wear full-face motocross style helmets. Such helmets have previously been shown to significantly reduce the frequency and severity of head and brain injuries resulting from bicycle crashes[5,6,7]. However, these helmets are generally much heavier than normal open face bicycle helmets (~300 g), with a mass typically between 900 g and 1700 g based on manufacturer claims[8]. Though the additional mass may not be an issue for adult riders, who are more physically mature, it may result in additional neck loading in younger, less developed riders as a result of increased helmet mass relative to body mass. Greater neck strength, allied to activating the neck muscles in readiness for impact, have been proposed to reduce an athlete's risk of concussion during a collision[9,10]. Riders with smaller and weaker necks are suggested to be more likely to experience greater translational and rotational displacements of the head following impact[11]. However, this relationship remains somewhat inconclusive.

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

Therefore, the aims of this study were to identify the number of accelerations, the magnitude of translational and rotational neck accelerations during BMX in different chronological age groups; to determine the influence of wearing a neck brace on these accelerations and to determine range of motion (ROM) with and without helmet and neck brace. The study also aimed to determine whether any relationships existed between the number of accelerations, magnitude of accelerations, rider age and helmet mass relative to body mass (RHM). It was hypothesised that neck accelerations would be greatest in younger riders and that the neck brace would reduce the magnitude of accelerations without affecting neck ROM. Finally, given the heavier RHM, it was hypothesised that relationships would exist between this and age, number of accelerations and the magnitude of accelerations.

2. Methods

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

The track was a national standard, indoor BMX track and had a 5 meter high start ramp with a 28° decent angle. Track length was 400 meters and consisted of three banked corners (berms) and four straight sections with a number of technical jumps on each straight. Riders performed three laps of the track without a neck brace (NB) and three laps with a neck brace (WB). The neck braces (Atlas, Atlas Brace Technologies, Valencia, USA) came in three sizes based on chest size (53-63 cm, 61-71 cm and 74-84 cm) and weighted 375 g, 460 g and 590 g, respectively. The manufacturers' guidelines for fitting of the neck braces was followed. This first required measuring the chest circumference at the level of the axilla and selecting the appropriate neck brace for that size. Secondly, the rear positioning mounts on the neck brace were adjusted to ensure the chest, back and shoulder pads sat flush against the body for each participant. As the neck brace was not directly attached to the helmet and was fitted to minimise movement around the neck and shoulder complex, it was deemed unlikely to contribute to translational or rotational accelerations of the head. Participants helmets were also weighed using a digital scale (Salter, Kent, UK) to the nearest 0.1 g, in order to determine RHM (g/Kg BM). A triaxial accelerometer with gyroscope (xPatch, X2 Biosystems, Seattle, USA) was used to measure the magnitude of translational (g) and rotational (rads/s^2) accelerations of the neck along with the number of accelerations for each trial. Sensors were positioned behind the right ear at the level of the occipito-temporal suture (Fig. 1). Separate sensors were used for the NB and WB trials for each rider. Translational accelerations were sampled at 1000 Hz, whilst rotational accelerations were sampled at 800 Hz. The minimum recording threshold was set to 5 g, whilst the sensors had a refresh rate of 100 Hz. The xPatch system had been validated previously for accelerations up to 160 g[14]. Any values recorded either above or below the minimum and maximum thresholds were deemed erroneous or 'clack' accelerations by the proprietary software (X2 Biosystems Injury Management Software) and removed from the dataset. As all riders were familiar with the track already, a 10 min warm up period was given prior to starting data collection. Riders were then instructed to ride full laps of the track as quickly as possible, without stopping, before returning to the start gate for a 5 min passive recovery between laps. The order of the trials were randomised and conducted over a three week period.

Range of motion (ROM) of the cervical spine, in all conditions, was assessed using simple 2D image processing. No participant reported any neck or spinale discomfort, nor had any musculoskeletal impingement at the time of measurement. A high quality digital camera (Nikon D5600) was mounted on a tripod, approximately 3 meters from the participant. In a seated position, with the head held in a neutral position, each participant performed three sequential flexion and extension movements. The head returned to the neutral position between each repetition. For the determination of ROM, post-processing of 2D images was undertaken using open-source software (ImageJ, https://imagej.nih.gov/ij/). With a perpendicular rule used for reference, an approximate line was plotted between the tragion and the orbitale. This was used to determine a neutral, or initial, angle. Deviation from this neutral angle, in both flexion and extension, was then calculated as the ROM, in all un-helmeted and helmeted conditions. ROM rotation measurements were determined in the supine position, again using a perpendicular rule for reference. Three sequential rotations to the right and left sides were completed. An approximate line was plotted between the bregma and nasal ridge. Again, deviation from the neutral angle, in both right and left rotation, was calculated as the ROM. For ROM movements, the average of the three measurements was calculated. To ensure objectivity, a second assessor, who was blinded to the previous measurements, also determined angles in post-processing. The coefficient of variation for cervical flexion was 1.6 %, extension 1.4 %, right rotation 2.1 % and left rotation 1.3 %. To determine intra-tester reliability, repeated measures were undertaken on two separate days. For all items, intraclass coefficients were $\geq$ 0.85. Assessment of lateral flexion was performed but not reported. This was due to the younger riders often being unable to maintain lateral flexion in the correct alignment when helmeted, potentially due to the increased mass of the helmet. As such, further kinematic analysis of the influence of helmet mass on cervical ROM should be advocated.

All data were analysed using the statistical software package SPSS (version 23 SPSS Inc., Chicago, IL). The alpha level was set at $p \leq 0.05$. Differences in accelerations and the number of accelerations between age groups and neck brace condition (NB vs WB) were determined using mixed model repeated measure ANOVA's (Condition x Age). *Post-hoc* analysis of within-subject effects were determined using a Bonferroni correction. Differences in RHM (g/kg BM) by age group were analysed using a one-way repeated measures ANOVA. Effect sizes were calculated using a partial Eta^2 (η_p^2). Effects sizes were identified as; small = 0.01, medium = 0.06 and large = 0.14[15]. Pearson's product moment correlations were used to determine any relationships between variables. Data are reported as mean ± SD (95 % CI) over the three laps for each condition unless otherwise stated.

Figure 1 near here

3. Results

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

Table 1 near here

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

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

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

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

Table 2 near here

4. Discussion

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

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

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

Despite this, it is of interest to note, that across all three age groups the translational accelerations observed were much greater than those previously reported for other sports. Lynall et al. (2016)[20] reported mean transitional loads of 12.51 g during collegiate level women's soccer, with the mean number of accelerations per 90 minutes of play ranging from 3.39 to 9.40, depending upon positional role. Participants in the present study experienced translational accelerations between 20.4 and 29.6 g, whilst the number of accelerations was more than double (6.4 to 17.5) those of the Lynall et al. study, yet in less than a 50 s period. Similarly, research into head accelerations in professional rugby league players also reported translational accelerations considerably lower (~15 g) than in the present study[21]. These findings demonstrate the scale of head accelerations during BMX riding, and in particular the severity of the accelerations with which young riders are exposed to. This may have implications for potential brain injuries and function. McAllister et al. (2014)[22], measured cognitive function along with using diffusion weighted imaging to determine brain white matter integrity and found both cognition and white matter integrity were impaired with repetitive impacts as low as 33.4g in soccer and ice hockey players. These impacts are comparable to those reported in the present study. Therefore, any means to reduce these accelerations should be welcomed by riders and governing bodies.

With respect to rotational accelerations, age did not significantly influence the magnitude of the accelerations. There was a significant difference between the NB and WB conditions though. However, whilst the use of a neck brace was shown to reduce the number of translational accelerations, the opposite was observed for rotational values. This is in opposition to the hypotheses. Though it is difficult to identify why the use of a neck brace would increase rotational accelerations, one possible explanation may relate to the riders perception of wearing them. Anecdotal conversations with the riders revealed the majority of them stated they felt it restricted their head movement. As such, it may be possible the riders overcompensated for the perceived limitation by consciously turning the head more when wearing the brace. This may have resulted in the higher rotational accelerations observed. However, further analysis is warranted to confirm this and to determine whether a learning effect would influence the results with greater practice with the neck brace. Our simple assessment of cervical range of motion revealed that a relatively consistent increase in range of motion, both translational and rotational, accompanied the wearing of a helmet, across all age groups. Peculiarly, for the very youngest group cervical flexion decreased when a helmet was worn. This is likely to be associated with the design of the full-face helmet, notably the pronounced chin area. This may have created a restriction on full range of motion when contacting with the upper sternoclavicular area. Introducing a neck brace did reduce range of motion in all cervical movements. This was expected, yet not significant.

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

5. Conclusions

This study found that BMX riders are exposed to high head accelerations regardless of age group when compared to other sports. Our findings show that the number of accelerations decreased with age, possibly as a result of muscular development about the neck and shoulders. It would also appear that the use of a neck brace could effectively further reduce the number of head accelerations across all age groups. However, the magnitude of these accelerations may be more related to riding dynamics and negative pre-conceptions relating to the wearing of neck braces. Lastly, RHM also appears to be influential in the number of accelerations observed. Therefore, the use of BMX helmets may place additional stress on the head and neck of younger riders.

Practical implications

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

Conflict of interest

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

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

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

Table 2. Mean ± SD (CI) number of accelerations, translational and rotational accelerations by age group over three laps.

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