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Biomechanical Investigation of the Efficacy of Conventional Bicycle Helmets at Preventing Facial Injury

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Abstract

Many studies indicate that wearing a bicycle helmet significantly reduces the risk of accident-related head injuries. Less is reported, however, about bicycle helmet use, about preventing facial injury. Epidemiological studies typically report incidences of 'head injury,' though rarely categorize head, brain, and facial injuries separately. Studies that do focus on facial injury suggest that helmets do not provide adequate protection. A 3D biofidelic child head form and two conventional bicycle helmets (that is," open face") were created in MSC ADAMS™ computer modelling and simulation software. Peak force and linear accelerations were measured for a series of head form impact simulations corresponding to an established testing standard and a newly proposed series of tests derived from modified motorcycle standards, incorporating injury thresholds from the facial bone impact tolerance literature. Almost all impact simulations exceeded the threshold for fracture, soft tissue, and dental trauma. An observational study was conducted using MSC ADAMS™ to assess a helmeted headform's face exposure during impact simulations. Helmets failed to reduce face impact exposure, during all simulations, except for perpendicular cheek impacts. International test standards, therefore, require urgent revision to ensure that face protection is included.

Publisher : Middle Technical University

Keywords: Biomechanics; Facial Injury; Bicycle-Helmet; Medical Instrumentation.

1. Introduction

Bicycle safety helmets are designed to prevent head injury, and although most commercially available helmets conform to established standards, individual designs may differ significantly. The effects of different designs on the ability of a helmet to provide face protection are currently unknown. There is emerging evidence that whilst the current design of bicycle helmets appears to prevent "head injury," there is an underlying failure to provide adequate facial protection. Indeed, a recent study [1] suggests that for a facial injury, bicycle helmets only provide a slight protective effect. Facial injury, however, appears to be particularly common, and its prevalence and links with dental trauma and brain injury have been noted in several studies [2-4]. Despite this, maxillofacial trauma is often overlooked, with more focus on head injury, which is considered more of a risk. When maxillofacial trauma is noted, it is often grouped alongside head injury in epidemiological studies. For example [2, 5, 6] concluded that approximately half of all victims of cycling accidents sustain some trauma to the head and face. It is, therefore, important that the differences between head and maxillofacial trauma be identified since the current bicycle helmet standards offer no requirements for facial protection. Whilst emerging evidence indicates that bicycle helmets serve to prevent "head injury," approximately one-half of all head/face injuries occur in regions that are not protected by the cycling helmet [6]. Cyclists incurring facial injuries have, historically, been relatively common, with dental trauma and brain injury being reported in several studies [2, 3, 4, 7]. As implied by Elvik et al. [1], however, there should continue to be investigation seeking a solution to decrease the incidence of facial injury in cyclists. In a detailed examination of the literature, Dickson et al. [9] reported that children at 10-year-old are at the most significant risk of experiencing a head injury following a cycling-related accident. A review of a series of helmets for children of such an age indicates that there is a range of products available [10], with the scientific community having evaluated the ‘traditional’ cycle helmet in some detail. Hence, this study will (1) employ computational simulation to evaluate the effectiveness of current, ‘traditional’ cycle helmets in mitigating the risk of facial injury. (2) The likelihood of injury will be assessed against the Abbreviated Injury Scale (AIS), a widely accepted technique that enables the prediction of injury when considering linear and rotational head acceleration data. (3) An observational study will be conducted to assess the helmeted headform’s face exposure during the impact simulations. (4) Evaluating the ability of a helmet to provide face protection from injury to revise the international test standards, to include face protection.

2. Materials and methods

Computer-aided design (CAD) software (Solidworks; Dassault Systèmes SolidWorks Corporation, MA, USA) was used to create models of a headform and two different cycling helmets. These models were then exported to a multi-body modelling software (ADAMSTM; MSC Software Corporation, CA, USA) to allow for the simulation and analysis of a series of standardised impact scenarios.

Nomenclature			
AIS	Abbreviated Injury Scale AIS	PCM	phase change material
CAD	Computer-aided design	μ	coefficient of friction
3D	Three Dimensional		

2.1. Headform model

A computational 'head form' with uniform cranial impact parameters was created that conformed to the appropriate British Standard to ensure the implementation of a consistent simulation methodology (BS: EN960; British Standards, 1997) [11]. Whilst this headform represented a 50th percentile male 10-year-old, some enhancements were performed to illicit an accurate, biomechanical response. Assigning appropriate material properties of the skull/cranium stiffness necessitated interpolation of similar results. Yoganadan et al. [12] report that by 6-8 years of age, the skull reaches approximately 75% adult stiffness [13-15]; therefore, based on the assumption that the skull reaches 100% stiffness by the age of 20yrs [25], linear interpolation was applied to find an approximate age-appropriate stiffness value for the headform model, see Table 1.

Table 1. Facial fracture tolerance values were tabulated from [22]

Bone	Force tolerance (N)	Pressure tolerance (N mm^{-1})
Zygoma	489-2401[16,17,18]	1.38-4.17 [18,20]
Zygomatic arch	890-1779 [20,21]	1.38-2.76 [17,20]
Mandible	685-1779 [16,17,18]	2.76-6.20 [21]
Maxilla	668-1801[17,19]	1.03-2.07 [20,21]
Frontal	1000-6494 [16,18,20]	≥7.58 [18]
Nose	342-450 [16]	0.13-0.34 [20]

It was assumed that the skull exhibits a uniform stiffness. Assignment of facial properties necessitated a range of stiffness values to reflect the variation in the properties across the regions of the face most susceptible to injury (i.e., chin (mandibular), midface (subnasal/maxillary), and cheek (zygomatic) regions). Interpolation of existing data was again required to approximate the properties of a 10-year-old child’s facial bones, see Table 2.

Table 2. Stiffness valued interpolated stiffness values for a 10-year-old

Facial Region	Average stiffness values (Nmm^{-1})	Interpolated stiffness for a 10-year-old (Nmm^{-1})
Chin/ Mandibular	475 [23]	392
Cheek/ Zygomatic	150 [24]	124
Sub-nasal/maxillary	120 [19]	99
Nasal	150 [24]	124

The parameter "penetration depth" enables the ADAMSTM software to model the flex and deformation of solid structures. Several studies report the thickness of the soft tissue of children's faces [26,27, 28]. Average thickness values for 9–10-year-olds were obtained for the chin and midface, corresponding to 9.8mm and 12.1mm, respectively. Measurements of the soft tissues of the nose and cheek for children of 10 years of age were undocumented. Data was, therefore, sourced from tissue measurements of fresh and embalmed cadavers taken using needle puncture [29]. An average thickness value of 7.1mm was obtained for the cheek of a 9-10-year-old. Values for the soft tissue of the nose could not be found; therefore, an assumption was made that during an impact, the soft tissues of the nose would fully deflect by a distance of 25mm. To simulate the effects in ADAMSTM software, a 'contact' was established with several parameters defining how the head form and helmets interact with the impact surface. The surface was modelled to represent a road as an abrasive anvil surface and was assumed to be flat and completely rigid. A frictional coefficient of 0.5 was adopted from another study, which defined the contact between the headform and anvil that was investigated to demonstrate the validity of the initial simulation [9]. To improve the biofidelity of the un-helmeted, Dickson et al. [9], head form, stiffness, penetration depth, and friction parameters were modelled. To define the coefficient of friction between the human head and the impact surface, it is obligatory to assess the literature; while different values of skin friction are available between 0.5-1 [9,25] [30-32], there is no description on the scalp or the hair interactions, therefore this study used two different fractional coefficients. Firstly, a value represented by $\mu = 0.7$, which is higher than the helmet-anvil frictional coefficients, replicating a 'worst case'. Secondly, a value represented by $\mu = 0.58$, which is approximately a mid-point between the ranges of frictional coefficients between the head-anvil and the helmet-anvil [9,25] [30-32].

2.2. Unhelmeted head form impact simulations

Bicycle helmets must currently comply with national safety standards. The bicycle helmet standard adopted by the UK is BS: EN1078 (British Standards 1997), which prescribes two perpendicular linear impact tests, with a 250g limit for linear acceleration. No standard, test, or corresponding limit currently exists for bicycle helmet face protection. Therefore, the British standard EN1078 [11], which defines accepted standards for ‘traditional’ cycle helmets, was applied to allow comparison and analysis of existing perpendicular impact studies. Both the un-helmeted and helmeted headforms were rotated to investigate the potential risk of perpendicular and oblique impact exposure to the facial regions of interest (chin, cheek, midface, and frontal bones, see Fig. 1. All impact simulations were performed using 1500 steps over 0.9 seconds to produce testing conditions consistent with BS EN 1078 [11].

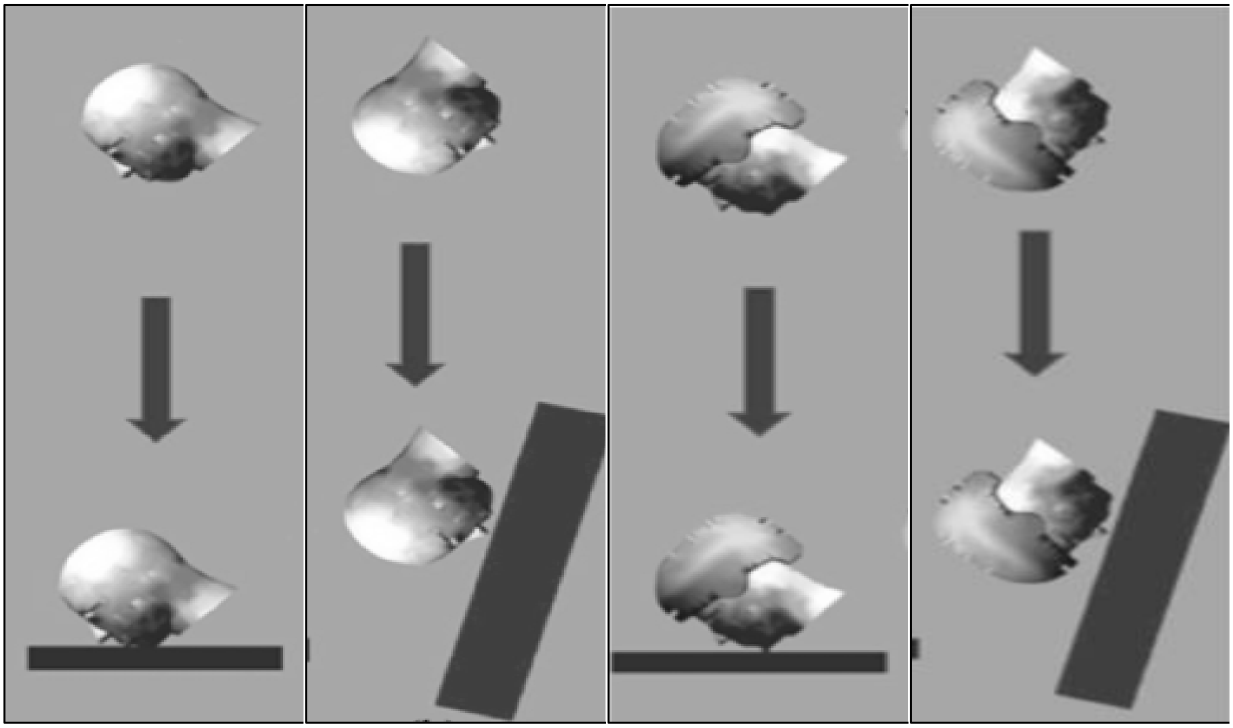


Fig. 1. Example simulation, showing contact between the chin and anvil, with headform unhelmeted (left) and helmeted (right), during perpendicular (a, c) and oblique (b, d) impacts

2.2.1. Perpendicular impacts

A series of perpendicular impact simulations were conducted with the un-helmeted and helmeted headforms (see Figs. 2 and 3), positioned to impact the anvil perpendicularly with the chin, cheek, midface, and frontal bone. The British standard EN1078 [11] was applied for perpendicular impacts to a plane surface at a velocity of 5.42ms^{-1}. The chin impact simulation, as shown in Fig. 2a, was described by British Standard BS: 6658 [33] and Snell CMR/CMS 2007 Snell/FIA Helmet Standard [34-37] (considered more suitable for children). The cheek (Fig.2(b)), midface (Fig. 2(c)), and frontal bones (Fig. 2(d)) were subsequently investigated. As standard tests do not currently consider facial impacts other than the chin, a new testing protocol was implemented that consisted of a series of simulated impacts corresponding to commonly injured facial regions and their relative biomechanical response parameters, as derived from the literature. In addition, a frontal bone impact simulation was conducted to investigate the relative risks from impacts to the face compared to impacts to the cranium. Frontal bone properties corresponded with Dickson et al. [9], that is, a stiffness of 471 Nmm^{-1} and penetration of 4.0mm, repeated for both friction values (μ=0.5 and 0.7).

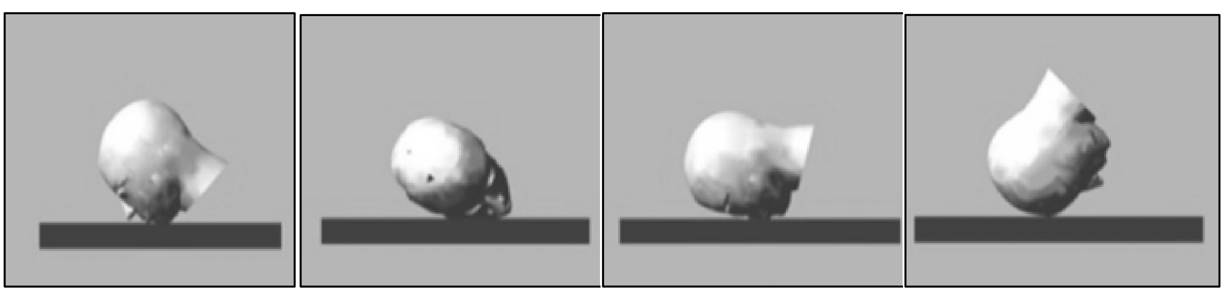


Fig. 2. Example showing the unhelmeted head form, from left to right, impacting the anvil perpendicularly with the chin (a), cheek (b), midface (c), and frontal bone (d)

2.1.2. Oblique impacts

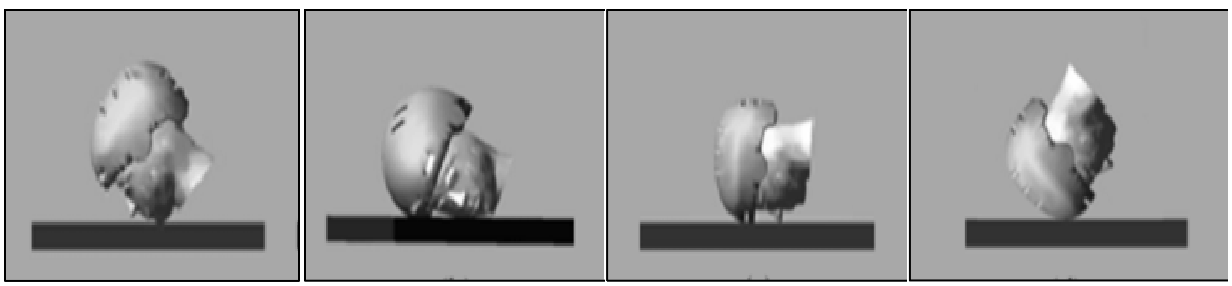
A series of oblique impact simulations were conducted (shown in Fig.1(b&d)) with the un-helmeted, and helmeted headforms positioned to impact an anvil, positioned obliquely concerning the chin, cheek, midface, and frontal bone. Oblique impact simulations were performed by a motorcycle test (UN-ECE Regulation 22.05 - Method: A) since BS: EN1078 [10] only includes perpendicular impact cases. The experiment aimed to reconstruct a fall vertical impact of 0.25m by 30km/h (8.2ms^{-1}) [25]. Likewise, this procedure is also used to investigate the helmet's performance [10]. This procedure has been interpreted to mimic a fall from a vertical plane at the height of 3.682m, resembling the value of 8.2ms^{-1}, onto a 15° inclined surface [25].

2.2. Helmeted head form impact simulations

Two bicycle helmets were selected from [10], based on their displayed advertisements in the market of their helmets, from their highest ('helmet 1') to lowest ('helmet 2') monetary costs. These helmets were then purchased before being manually measured and then drawn in the above CAD software (i.e., 'reverse-engineered). The helmet models were positioned in a manner relative to the headform, relying on the fitting guidelines of the manufacturer. Between the helmet and head form, a void was formed, which was then filled with padding. During impact, the helmet was constrained to the headform rigidly to characterize the zero yield (worst-case) between the chinstrap and chin, while between the helmet and head form used a zero slippage. The friction coefficient between each helmet and the ground was assigned as μ=0.58 for 'helmet 1' ('hard shell helmet') and μ=0.66 for 'helmet 2' ('soft shell helmet') [10]. The construction of the headform-helmet was validated against published experimental data [10]. For the correlation of the headform between the computational and experimental test, the mass of the head form was raised to 4.2kg, and the inertial properties were amended using documented data by Reynolds et al. [38]. The impact simulations were performed (a) on the side region of the helmet using 5.42ms^{-1} (1.5m fall height) and (b) on the helmet front rim using 4.57ms^{-1} (1.0m fall height), replicating the nose down. By comparing the measured acceleration to the published data and modifying the stiffness of the helmet iteratively, validation was achieved. From the simulation test, the proper contact properties were verified for the head and helmet, so the impact simulations of the helmeted headform of 10 years old were performed after reducing the head mass to that specified age. Since this study was conducted to assess the likely impact of exposure to the face during conventional (that is, 'open face') helmet use, a further series of simulations were conducted by the Standard (British Standard BS: 6658 and Snell CMR/CMS 2007 Snell/FIA Helmet Standard) and tolerance data in the literature. Simulations were conducted as above, both perpendicularly and obliquely (see Fig.1). Facial impact exposure was noted in the 3 facial regions of interest, the chin, midface, and cheek regions, and the front of the helmet (see Fig. 3).

2.2.1. Perpendicular Impacts

A series of perpendicular impact simulations were conducted with the helmeted headform positioned precisely as in the unhelmeted simulations (see Fig. 3) to impact the anvil perpendicularly with the chin, cheek, midface, and front of the helmet. Initially, the un-helmeted head form was positioned at an angle required to produce contact with the required area of the face. After the unhelmeted simulation had been run, the helmets were added and locked to the headform using the padlock joint in AdamsTM. The chin impact simulation was conducted as described by British Standard BS: 6658 [33] and Snell CMR/CMS 2007 Snell/FIA Helmet Standard [34-37] (see Fig. 3a). Exposures of the cheek (Fig. 3b), midface (Fig. 3c), and frontal bones (Fig. 3d) were investigated, such that the helmet could be assessed for its ability to provide face protection.



(a) (b) (c) (d)

Fig. 3. Example showing the helmeted head form, from left to right, impacting the anvil perpendicularly with the chin (a), cheek (b), midface (c), and front of the helmet (d)

2.2.2. Oblique impacts

A further series of oblique impact simulations were conducted to investigate face exposure (shown in Fig. 1d) with the helmeted headform positioned to impact the anvil obliquely with the chin, cheek, midface, and front of the helmet. Oblique impact simulations were performed, with an impact velocity of 8.5ms^{-1}, simulated by dropping the helmeted head forms from a height of 3.68m.

3. Results

3.1. Unhelmeted head form impact simulations

3.1.1. Perpendicular impacts

A series of head forms for perpendicular impact simulations were conducted to recreate the British standard EN1078 bicycle helmet test [11], with an effect onto a hard surface (anvil), positioned at 90 degrees to the direction of travel at 5.42ms^{-1}. The impact-induced accelerations at the chin, cheek, and midface were generally below the prescribed maximum British Standard bicycle head threshold of 250g, for all tests, except the cheek, with a skin friction coefficient of μ=0.7, see Fig. 4. However, even though the simulations fell within the 250g British Standard threshold, the minimum accelerations corresponded to at least moderate injury (AIS 2); 50% of the impacts (6 of 12) produced accelerations greater than 200g, corresponding to severe injury (AIS 3). Impacts on the chin had acceleration values of 233 and 234g (for frictional coefficients of μ=0.5 and μ=0.7, respectively), corresponding to an AIS 4 'severe' injury. Surprisingly, the acceleration for the cheek impact, with skin friction of μ=0.7, was 319.6g, corresponding to an AIS 6, classed as 'un-survivable' by St Clair and Chinn [10]. From Fig. 5, The contact forces produced during the unhelmeted perpendicular impacts with corresponding upper and lower fracture tolerances were represented by 17.0kN and 17.1kN at the chin; 11.3kN and 16.7kN at the cheek, and 10.8kN and 10.9kN at the midface, for the friction coefficients of μ=0.5 and μ=0.7 respectively. The face impact forces were lower than the frontal bone impact forces represented by 19.0kN and 18.6kN for the friction coefficients of μ=0.5 and μ=0.7, respectively. The simulated perpendicular face impacts produced contact forces significantly over 5kN, corresponding to a high fracture risk compared to the fracture threshold for the three impacted facial bones, which was less than 2.5kN as reported by [22].

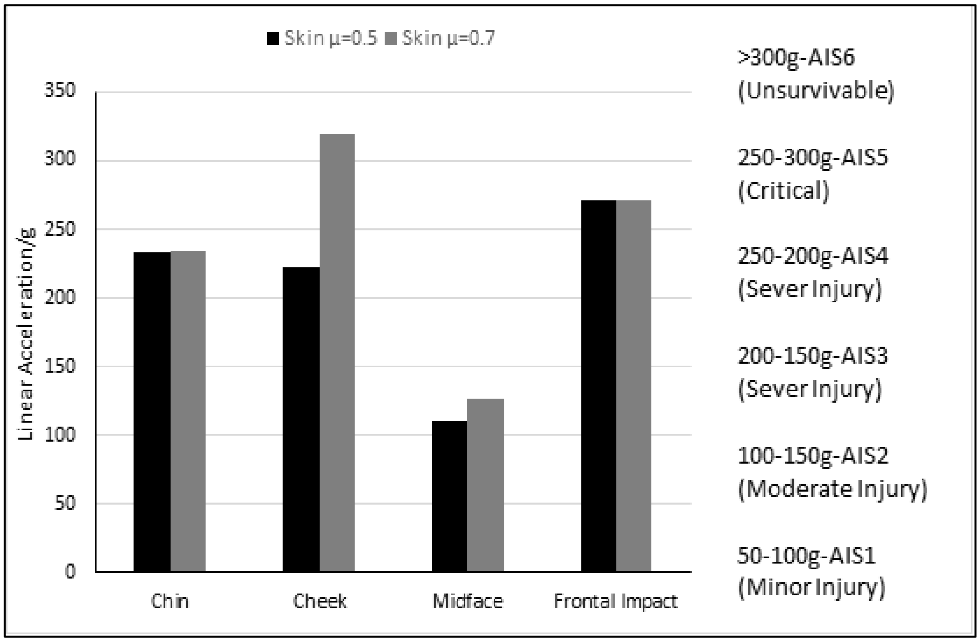


Fig. 4. Linear accelerations during unhelmeted perpendicular impacts to the chin, cheek, midface, and frontal bone with corresponding Abbreviated Injury Scale (AIS) scores

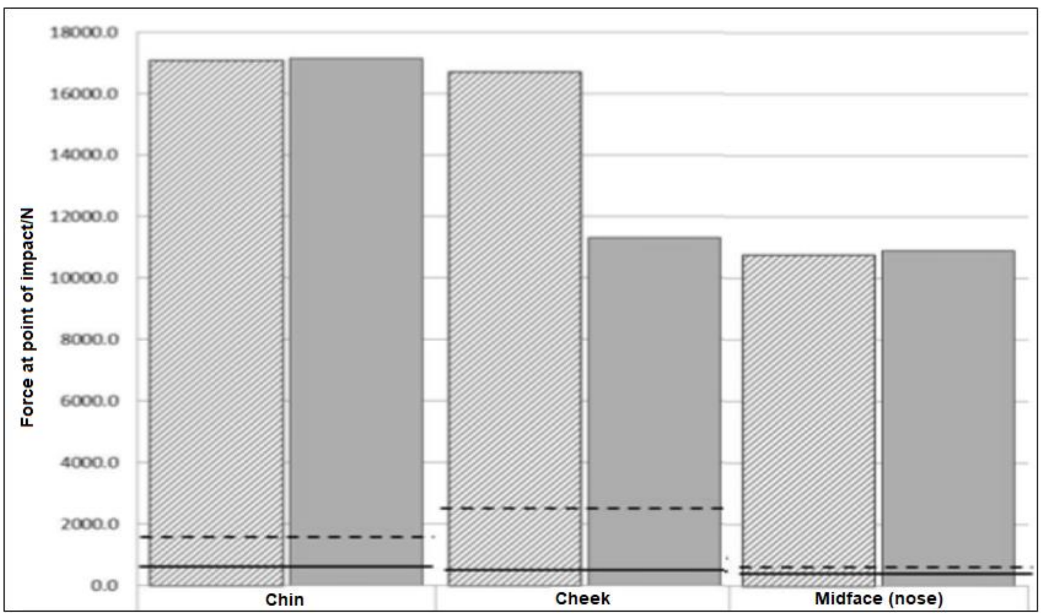


Fig. 5. Forces produced during unhelmeted perpendicular impact to the chin, cheek, and midface, with corresponding upper (-----) and lower (____) fracture tolerances

3.1.2. Oblique impacts

A series of headform oblique impact simulations were conducted by a motorcycle test, with an impact onto a hard surface (anvil), positioned obliquely concerning the chin, cheek, midface, and frontal bone to the direction of travel at 5.42ms^{-1}. From Fig.6, impacts on the headform, with a skin coefficient of friction of μ=0.7, produced peak accelerations ranging from 93g at the chin to 82g at the midface. The head form with a lower skin coefficient of friction of μ=0.5 produced lower accelerations ranging from 87g at the chin to 75g at the midface. All oblique impacts to the un-helmeted headform were in the range of an AIS 1 level injury, corresponding to minor injury [10]. Interestingly no impacts were severe enough to cause AIS 2 injury, compared to the tests performed by Dickson et al. [9], where the un-helmeted headforms experienced accelerations of a minimum of 122g, corresponding to an AIS 2 injury. The oblique impact simulations produced lower linear accelerations than the perpendicular impacts, see Fig. 6. From Fig. 7, The contact forces produced during the unhelmeted oblique impacts with corresponding upper and lower fracture tolerances were represented by 3.15kN and 3.38kN at the chin; 2.38kN and 2.31kN at the cheek, and 2.74kN and 2.97kN at the midface, for the friction coefficients of μ=0.5 and μ=0.7 respectively. The oblique impact contact forces were significantly lower than the perpendicular impacts. The maximum oblique impact force, sustained by the un-helmeted head form, was 3.4kN compared with the maximum perpendicular impact force of 17.1kN. From Fig. 8, Also the oblique impacts to the head forms produced angular accelerations for impacts at the chin, cheek, and midface, see Fig. 8. The impact on the chin produced a peak angular acceleration of 10679.9 rads^{-2} for μ=0.5 and 14721 rads^{-2} for μ=0.7. The effect on the cheek produced 10792.2 rads^{-2} for μ=0.5 and 11642.1 rads^{-2} for μ=0.7. These outcomes correlated with a risk of a fatality compared to the lower angular accelerations produced by the midface impact represented by 767.1 rads^{-2} for μ=0.5 and 1079.4 rads^{-2} for μ=0.7 which is below the limits reported by [39] and [40].

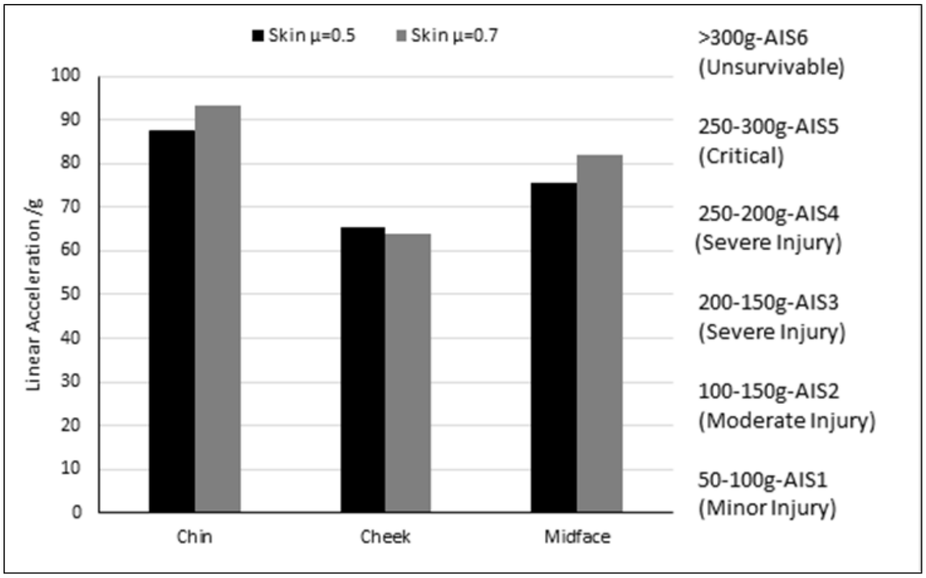


Fig. 6. Linear accelerations during unhelmeted oblique impacts the chin, cheek, and midface with corresponding Abbreviated Injury Scale (AIS) scores

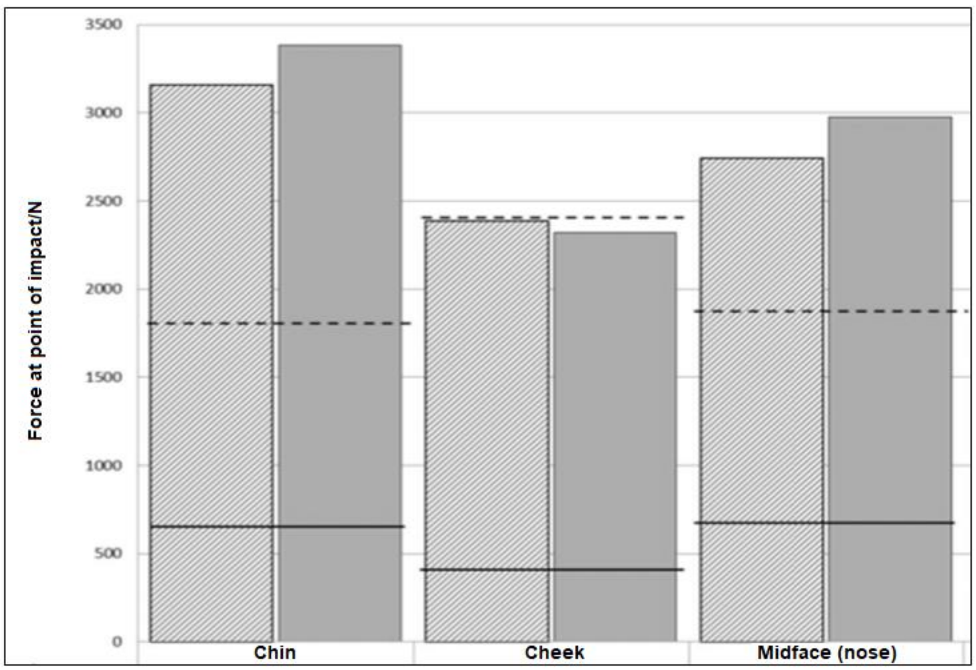


Fig. 7. Forces produced during unhelmeted oblique impacts to the chin, cheek, and midface, with corresponding upper (-----) and lower (____) fracture tolerances

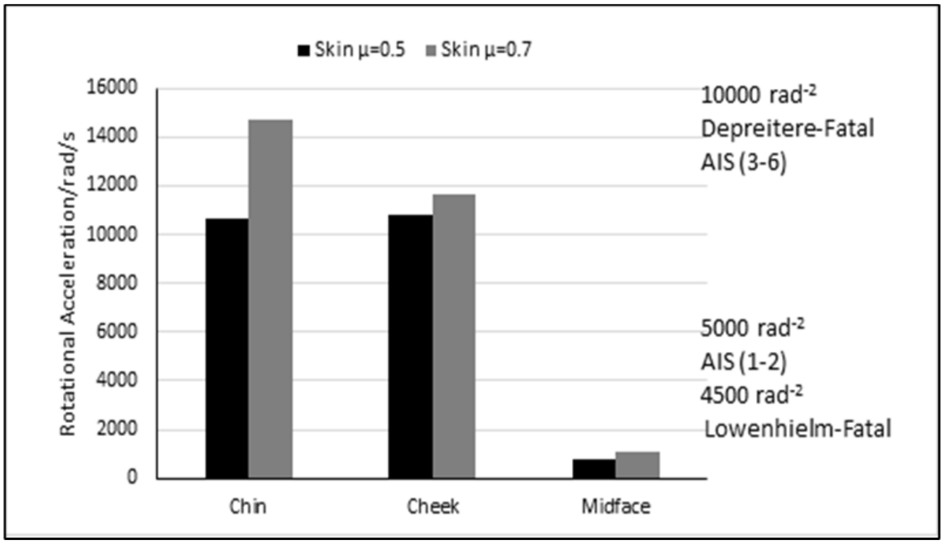


Fig. 8. Angular accelerations about the centre of mass after oblique impact, with corresponding Abbreviated Injury Scale (AIS) scores

3.2. Helmeted head form impact simulations

3.2.1. Perpendicular impacts

The perpendicular helmeted headform impact simulations were conducted by the methodology described above (at 2.3.1.). Observations were made of face contact exposure during impact simulations, as shown in Table 3.

Table 3. Face contact exposure as a result of the perpendicular helmeted headform impact simulations

Perpendicular impact	Chin	Cheek	Midface
Helmet 1	Face contact	Face contact	Helmet contact
Helmet 2	Face contact	Face contact	Face contact

3.2.2. Oblique impacts

A series of oblique helmeted headform impact simulations were conducted by the methodology described above at (2.3.2). Observations were made of face-contact exposure during simulated helmeted headform simulations, as shown in Table 4.

Table 4. Face contact exposure as a result of the oblique helmeted headform impact simulations

Oblique impact	Chin	Cheek	Midface
Helmet 1	Face contact	Face contact	Face contact
Helmet 2	Face contact	Face contact	Face contact

4. Discussion

A 3D biofidelic child head form and two commercially available 'open face' bicycle helmets were created in MSC ADAMS™. Peak force and linear accelerations were measured for a series of head form impact simulations corresponding to an established testing standard and a newly proposed series of tests derived from modified motorcycle standards, incorporating injury thresholds from facial bone impact tolerance literature. A further series of perpendicular and oblique head form impact simulations were conducted with the two commercially available 'open face' bicycle helmets, and the relative face exposure was noted.

4.1. Unhelmeted head form impact simulations

4.1.1. Perpendicular impacts

Current regulations appear to focus on cranial impacts, disregarding the potential risks of face impact-induced linear acceleration producing brain injury [41-43]. To compare the relative risks of effects to the cranial bones and cranial soft tissues with the comparatively less stiff facial bones and more compliant soft tissues, this study simulated a frontal bone impact. Frontal bone impact accelerations of 270 and 280g were produced for friction coefficients of μ=0.5 and 0.7, respectively, as shown in Fig. 4.

About the contact forces experienced by the un-helmeted head forms, forces of 17.0kN and 17.1kN were produced at the chin; 11.3kN and 16.7kN at the cheek, and 10.8kN and 10.9kN at the midface, for the friction coefficients of μ=0.5 and μ=0.7 respectively. The forces produced at the face were lower than those sustained at the frontal bone impacts (19.0kN and 18.6kN), further supporting the suggestion that the facial structures may provide a protective effect to mitigate neurological injury.

The fracture tolerances see Table 1 from [21] for the three impacted facial bones were less than 2.5kN, as shown in Figs. 5 and 7. The simulated perpendicular face impacts shown in Fig. 5 produced contact forces significantly over 5kN, corresponding to high fracture risk. The high forces, more than the fracture thresholds, suggest that if a fracture did occur, it would be likely to be severe. Melvin and Shee [44] noted that only the nasal bone fractured at forces less than 6.2kN when cadavers were impacted against a large plate. This was attributed to the facial bones working together to dissipate forces through the struts of the face. However, since all the forces produced by the un-helmeted headforms were more than the 6.2kN threshold (minimum of 7.8kN), fracture risk must be considered high. The high values suggest that even though children may have a higher fracture tolerance [48], the fracture is still likely to occur, though maybe at a lower severity than that sustained by adults. Although the maxilla was seen to sustain the lowest force, as shown in Fig. 5, Lieger et al. [46] concluded that patients with a fracture of the mandible were most likely to have a dental injury (39%). The higher the contact forces experienced by the mandible, the greater the risk of fracture being sustained to all parts of the face and the greater the risk of dental trauma. It is acknowledged that the perpendicular impact tests, prescribed by EN1078, cannot be considered entirely realistic since it is highly unlikely that a bicyclist would impact an object entirely perpendicularly, absent of any force being dissipated in other directions, as noted by Otte [47].

4.1.2. Oblique impacts

The oblique impact simulations produced lower linear accelerations than the perpendicular impacts, see Fig.6. This was expected, since the headform was observed to impact the anvil at a glancing angle, with the initial impact creating the greatest acceleration and force. Impacts on the headform, with a skin coefficient of friction of μ=0.7, produced peak accelerations ranging from 93g at the chin to 55g at the midface. The head form with a lower skin coefficient of friction of μ=0.5 produced lower accelerations ranging from 87g at the chin to 51g at the midface. All oblique impacts to the un-helmeted headform were in the range of an AIS 1 level injury, corresponding to minor injury [10]. Interestingly no impacts were severe enough to cause AIS 2 injury, compared to the tests performed by Dickson et al. [9], where the un-helmeted headforms experienced accelerations of a minimum of 122g, corresponding to an AIS 2 injury. This reduction in linear acceleration can be attributed to both: the less stiff facial bones, which are more than 4 times less stiff (99 - 392 Nmm^{-1}), than the skull stiffness value (471 Nmm^{-1}), and a greater facial soft tissue penetration depth than used by Dickson et al. [9]. The oblique impact contact forces were significantly lower than the perpendicular impacts, the maximum oblique impact force, sustained by the un-helmeted head form, was 3.4kN compared with the maximum perpendicular impact force of 17.1kN.
The forces sustained by the un-helmeted head form, for impacts to the chin and midface as shown in Fig.7, correspond to a high risk of fracture, being significantly greater than the tolerances quoted by Hampson [22] from Table 1. Conversely, the impacts to the cheek were within the fracture tolerance band and therefore reflect a lower risk of fracture. This may be particularly relevant in children, who are reported to have higher fracture tolerances [38]. However, in the simulations, shown in Fig.7, where forces were less than 6.2kN, no fractures would be expected [47]. The greater forces at the mandible provide a possible explanation for the high numbers of mandible fractures sustained by bicyclists, with approximately two-thirds of these being condylar. In the perpendicular impacts, the high risk of fracture is associated with a correspondingly high risk of dental trauma, with the highest forces again occurring at the mandible, where most fractures display concurrent dental trauma [40]. In addition, the oblique impacts to the un-helmeted headforms produced angular accelerations for impacts at the chin and cheek, which by Löwenhielm et al. [39] and Depreitere et al. [40] could be associated with a risk of a fatal outcome. The impact to the chin produced a peak angular acceleration of 14721 rads^{-2}, significantly above the limit of 4500rads^{-2}, for fatality [39] and 10000rads^{-2} [40]. The midface conversely created relatively low accelerations below all limits (767.1 rads^{-2} for μ=0.5 and 1079.4 rads^{-2} for μ=0.7). The midface impact accelerations were low, compared to the chin and cheek impacts, since the headform contacted at an angle, such, that instead of rebounding from the surface, it skidded across it.

4.2. Helmeted head form impact simulations

A further series of perpendicular and oblique head form impact simulations were conducted with the helmeted head form. The two commercially available 'open face' bicycle helmets were added to the head form and the simulations, described above, were repeated; face impact exposure was noted as shown in Tables 3 and 4.

4.2.1. Perpendicular impacts

Chin and midface contacts were observed during head form impact simulations with both helmets '1' and '2'. No helmet contact was observed during any of the chin and midface simulations. During perpendicular impacts to the cheek, however, helmet '1' was observed to contact 5msec before the cheek. Thus, the risk of injury to the chin and midface was unchanged by helmet use. About the cheek, helmet '2' provided no protection; however, helmet '1' did act as the primary impact site and can therefore be considered to provide some protection.

4.2.2. Oblique impacts

During the chin impact simulations, the chin was again observed to be the primary contact surface with the anvil, neither 'helmet 1' nor 'helmet 2' was observed to provide any protection. The midface impacted first at the nose and then rotated, such that the helmet impacted the anvil, this, however, was due to the solid body contact model limitation of the nose, rather than a representation of reality. It can be inferred, however, that the midface was unprotected during the simulation. Regarding the cheek impacts, the cheek was observed to impact the anvil independently of both 'helmet 1' and 'helmet 2'. Thus, both helmets '1' and '2' failed to provide any head form face protection, such that during all the simulated impacts there was an unchanged risk of fracture, soft tissue, and dental trauma.
The limitation of the study was the nonmodelling of the neck, this study used the head form only that represents the effective head's mass as reported by [49,50,51]. Based on [49], modelling the head produces (1) a primary acceleration time response subsequent by the influence of the neck producing (2) a secondary acceleration time response comparatively long period. To preclude the head-neck coupling complex response and to determine the impact of the head isolated only the head was simulated in this study.

5. Conclusions

This study explored the ability of current, 'traditional' cycle helmets to mitigate the risk of facial injury. Peak force and linear accelerations were measured from a series of 3D biofidelic child headform impact simulations, incorporating injury thresholds from facial bone impact tolerance literature. The measured values for all impact simulations exceeded the threshold for fracture, soft tissue, and dental trauma. The headform was helmeted and the simulated tests were repeated. An observational study was conducted to assess the helmeted headform's face exposure, during the impact simulations. Helmets failed to reduce face impact exposure, during all simulations, except perpendicular cheek impacts, where one helmet was observed to be the primary impact site. From this study, it was concluded that the international test standards need important revision to make sure that face protection is included.

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