5 Biomechanical Studies of Impact and Helmet Protection

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Introduction

Definitions of concussion have evolved over time and these definitions should inform our interpretation of past research. Studies on concussion in the not-too-distant past may have examined a constellation of brain injuries that are more severe than those currently considered as sport-related concussion (SRC). The current definition and signs and symptoms of SRC have been informed substantially by consensus statements [1]. SRC is defined as the outcome of a biomechanical load applied to the head directly or indirectly. Helmets have a well-proven role in managing loads applied directly to the head. However, we have been more successful in developing helmets to prevent moderate-to-severe head injuries, rather than SRC. For example, in August 2020, Riddell, a major supplier of helmets for American football, warned: "Contact in football may result in CONCUSSION-BRAIN INJURY which no helmet can prevent".

Developing effective helmets for sport is challenging. Intrinsic and extrinsic factors and the exposure profile of the inciting event all require consideration and realization in an affordable, lightweight and comfortable device that does not impede athletic performance or enjoyment. Intrinsic risk factors include age, gender, injury history, anatomy and behaviour. Extrinsic risk factors include the laws and rules of the game (especially around head contact), the environment (e.g. the playing surface from soft ground to ice) and the use of personal protective equipment and/or coaching strategies. The inciting event might be summarized into a small predictable pattern (e.g. in soccer head-to-head or arm-to-head impacts during the aerial contest

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for the ball, also known as "elbowing"), be broad, or even unknown. Through the use of video and wearable head impact sensors, knowledge is being gained regarding exposure profiles across many sports and levels of play. In the context of SRC, this chapter describes current knowledge regarding helmet performance, consider helmet design characteristics and standards, human factors, research and development needs, and opportunities. A focus of the chapter is on padded or softshell headgear that is worn in sports such as rugby union, Australian football and combat sports.

Epidemiological Approaches – Effectiveness and Efficacy

Epidemiological studies in sport show that at present helmets cannot be relied upon as the primary method to prevent concussion [2–4]; Table 5.1. In a sporting team or organization, it is not possible to satisfy a duty of care by mandating helmet use. In some sports, for example, Australian rules football, rugby union, rugby league and soccer, there is no evidence that helmets, referring to padded headgear, may prevent concussion. In American football and ice hockey, the epidemiological evidence regarding the benefits of helmets in preventing concussion is inconclusive. In both these sports, there is inconsistent evidence that helmets are effective in preventing head injuries overall.

One of the major impediments to the use of epidemiological methods to assess the role of helmets in sports that have mandatory helmet use, for example, American football, is that comparisons can only be made between types of helmets, not

Table 5.1 Summary of effectiveness of helmets in preventing concussion

Sport	Concussion rate (games)	Proportion of injuries (%)	Helmet mandatory	Effective in reducing concussion
Rugby union	4.1–6.9 per 1000 player hours (all levels)	5–15	No	No
American football	0.5–5.3 per 1000 athletic exposures (high school and collegiate)	5	Yes	Inconclusive
Football (soccer)	0.06–1.08 per 1000 player hours	3	No	No
Ice hockey	0.2–6.5 per 1000 player hours (collegiate and professional)	2–19	Yes, including face shields in some competitions	Inconclusive
Bicycle riding	Not quantified	Depends on sample inclusion criteria	City, state and country dependent	Yes

There are variations in injury rate measures based on injury definitions, exposure measurements, chosen denominator, level of play and age groups assessed between athletes assigned randomly to a helmet group and a no helmet group. In 2013, McGuine's study reported no difference in concussion risk by helmet brand or year of manufacture amongst high school football players [5]. In an earlier study, Collins observed that a smaller proportion of high school football players wearing the then new Riddell Revolution® helmet were concussed (5.3%) than players wearing standard helmets (7.6%) [6]. A comparison between players wearing and not wearing a helmet is not possible. Thus, the overall benefit remains unclear. To this end, one American football helmet manufacturer advises the public that: "Scientists have not reached agreement on how the results of impact absorption tests relate to concussions. No conclusions about a reduction of risk or severity of concussive injury should be drawn from impact absorption tests", "No helmet system can prevent concussions or eliminate the risk of serious head or neck injuries while playing football", and "No helmet system can protect you from serious brain and/or neck injuries including paralysis or death. To avoid these risks, do not engage in the sport of football" [7].

Other issues, for example, non-compliance, confound the conduct, results and analysis of epidemiological studies. Non-compliance may arise in sports where helmet use is not mandatory and athletes are randomized to a helmet-wearing group but do not normally wear a helmet. In the largest randomized control trial of helmets in sport, the author and colleagues found actual helmet wearing compliance to be poor in each of the three study arms, which may have weakened the positive trend observed with the "modified" helmet for those players who stuck with wearing the helmet during the study [8]. In a compliance analysis, wearers of the "modified" headgear compared to non-wearers had a non-significant reduction of greater than 50% in the likelihood of concussion causing one missed game. Players reported that the "modified" helmet, which was thicker and heavier than the "standard" design, felt stiff and uncomfortable. Although helmets in rugby union are substantially lighter than in American football, the perception relative to the experience of an even lighter headgear or no headgear influenced compliance.

Bicycle helmets have been shown to reduce the likelihood of concussion when the injury patterns of helmet wearing bicycle riders are compared to non-wearers. In an analysis of admissions to a major metropolitan trauma centre bicycle riders wearing helmets were observed to have a 54% reduction in the likelihood of concussion and a 66% reduction in the likelihood of intracranial injury (including concussion) compared to bicycle riders not wearing a helmet [9]. In bicycle crashes with motor vehicles, a training hazard for professional and recreational sports cyclists, the majority of brain injuries (79%) were considered concussive or involved loss of consciousness [10]. Moderate concussive injuries were associated with a 46% reduction if a helmet was worn. Concussion cases in trauma admission data may be based on different diagnostic criteria, for example, the International Classification of Disease (ICD) or the Abbreviated Injury Scale (AIS), than those in many helmet studies in football where the sports concussion consensus guidelines have been applied.

Helmet Characteristics

The most important functional characteristic of a helmet in the context of concussion is impact energy attenuation; a characteristic that has also been referred to as acceleration management. Ideally, the impact will cause the helmet to deform a substantial proportion of its thickness, without fully deforming or "bottoming out". The liner of the helmet or, the entire helmet in the case of padded headgear worn in rugby union, largely determines the impact energy attenuation performance. In short, the greater the deformation of the helmet, the greater the reduction in impact force as well as in head acceleration. The helmet can also distribute the impact force over an area larger than the contact area. In helmets with a well-established role in transport and sport, for example, bicycle, equestrian and motorcycle helmets, the helmet is designed for a single crash event. In contrast, American football, rugby union and ice hockey helmets are intended to provide protection throughout a season or more of multiple head impact exposures. The general properties of helmets and their function have been addressed well by many authors, for example, Newman [11] and Hoshizaki and Brien [12].

The next most important functional characteristics of the helmet are the mass, mass distribution, fit, restraint system, vision and thermal comfort. Sports helmets need to be wearable during extreme physical activities; therefore, helmet mass must be minimized. The mass distribution of the helmet and attachments is important in reducing the flexion moment that the helmet may apply to the head and neck. A flexion moment will be counteracted by neck extensor activation leading to muscle fatigue and increased joint reaction forces. It is imperative to ensure that the helmet and all components are correctly selected and adjusted for the individual athlete. Providing a kit bag with a few helmets to fit all the team is not the best practice. Vision and the restraint system characteristics are usually addressed in sports helmet standards. Where faceguards (visors) are mounted to helmets to prevent projectile to face or head impacts, the adjustment of the faceguard is important as apertures may permit a projectile travelling at speed to strike the face directly. In recent history, the position of a faceguard or visor on a cricket helmet could be adjusted by the player. As a result, there was potential for injury due to misuse. Current cricket helmets have a fixed position mounting for the faceguard. Therefore, positive changes are possible.

Performance Requirements and Standards

Helmet performance is assessed in the laboratory by examining the capacity of the helmet to minimize headform acceleration in impact tests. These tests are conducted against set criteria, for example, a linear acceleration pass criterion, or to derive an injury risk estimate. During a test, a selected amount of impact energy is delivered to the helmet-headform system via a drop rig, pendulum or mechanical device. The headform's linear and, in some cases, angular acceleration is measured during the impact. The input characteristics of the tests, for example, energy, dimensions of impact interface and headform, have developed to reflect knowledge on impact exposures in specific sports. The output characteristics, for example, headform dynamic responses, have also developed to reflect knowledge on injury mechanisms and human tolerance. However, requirements in many helmet standards are not currently aligned to maximize the potential for standard compliant helmets to prevent concussion. This would require the lowering of pass levels, for example, headform acceleration, to well below 100 g, and consideration for angular acceleration criteria and related test methods. As will be presented in this section, more valid assessments of helmet performance are observed when the laboratory tests reflect the impact exposures in the specific sport (impact location, impact severity, interface characteristics and frequency) and the biofidelity of the head-neck system are considered. A range of headforms are used in research and standards testing: Hybrid III headforms, rigid ISO headforms and NOCSAE headforms. Each has a distinct influence on the test outcomes. In an otherwise equivalent impact, head acceleration will be greater with a rigid ISO headform in comparison to a Hybrid III headform.

The author and colleagues have conducted baseline tests on bare headforms. These reveal a clear risk of concussion related to linear head acceleration even in impacts equivalent to the head falling 0.5 m (3.13 m/s):

- Hybrid II dropped onto a flat rigid anvil at 3.13 m/s has a peak linear acceleration (PLA) of 282 g and head injury criterion of 906 [13].
- Projectile impacts (ice hockey puck, baseball and cricket ball) into a Hybrid III headform mean PLA were in the range of 233 to 316 g for 19 m/s impacts and 342 to 426 g for 27 m/s impacts [14].
- Hybrid III headform mean PLA in flat rigid anvil was in the range 241 to 261 g (HIC 493 to 741) at 3.13 m/s and 368 to 512 g (HIC 1620 to 2789) at 4.43 m/s [15].
- Hybrid III headform PLA and peak angular accelerations (PAA) were measured in 16 linear impactor tests at five sites at a speed of 4 m/s. The impactor mass was 4 kg with a Polyurethane 70A Duro (Shore hardness 65 to 70) head. The average PLA = 140 g (SD = 17 g) and PAA = 8400 rad/s^2 (SD = 2100 rad/s^2).

In the context of laboratory impact tests, helmets need to reduce both linear and angular headform acceleration. As a guide, the 15% likelihood of concussion for adult males is 45 g and the 50% likelihood is 75 g for resultant linear acceleration at the head's centre of gravity [16]. For the bare headform impacts described above helmets need to reduce the linear acceleration in the range of two to tenfold to prevent concussion. Angular acceleration tolerance thresholds vary; Rowson reported that the 75% likelihood of concussion for resultant angular acceleration is 6.9 krad/s^2 [17].

Testing of a commercially available padded headgear model in Australia under the conditions described above (4 m/s linear impactor tests with 4 kg mass) showed a large reduction in PLA with one model, average PLA = 70 g (SD = 12 g, n = 15) and PAA = 4600 rad/s^2 (SD = 700 rad/s^2), demonstrating a potential to reduce head accelerations to a level suggestive of a protective effect in an equivalent severity impact.

How Well Do Helmets Perform?

Rugby Padded headgear in rugby must comply with World Rugby's performance regulations [18]. The helmet properties are restricted to an undeformed thickness of 10 mm and a foam density of 45 kg/m^3. World Rugby's impact performance requirements state that in a 13.8 J rigid (EN 960) headform impact onto a rigid flat anvil the peak headform acceleration shall not be less than 200 g. The mandated performance requirements exclude headgear from preventing concussion due to the biomechanical criteria and are inconsistent with the philosophy of many helmet standards.

Impact tests on helmets meeting World Rugby's requirements ("standard") and a "modified" version were conducted by the author [19]. The modified headgear was 16 mm thick and made from 60 kg/m^3 polyethylene foam. The standard headgear was 10 mm thick and made from 45 kg/m^3 polyethylene foam. Tests using a rigid headform from a 0.3 m drop height produced peak accelerations in the range 276–689 g for standard headgear and 69–123 for modified headgear. At 0.4 m peak, accelerations for the modified headgear were 110–273 g. The performance of the modified headgear in laboratory tests identified a potential in low severity impacts for the headgear to reduce the linear acceleration to a tolerable range. In the epidemiological study, there was a greater than 50% non-significant reduction in missed game concussions based on a compliance analysis [8]. With greater compliance, this may have been a significant association.

Figure 5.1 shows the results of linear impactor testing of a range of more recent (2016/2017) padded headgears marketed for use in Rugby Union, Rugby League and Australian football superimposed onto PLA-based injury likelihood curves [16]. The linear impactor was similar to that described earlier, but with a different impactor head. The results showed the little benefit of then current commercial models and large potential benefit of prototype models with respect to bare headform tests.

Australian Football Data on head impact exposures in Australian Football have emerged over the last few years. We undertook studies of player cohorts using a combination of video and x-patch sensors to measure head impact exposures in unhelmeted players [20, 21]. One of the aims of this research has been to assist in the development of standards. Setting aside the challenges and disappointments with the x-patch sensors, we observed:

- In 53 male and female community-level players (mean age = 26 years), there were 118 head acceleration events (HAE) with PLA $\geq$ 30 g, 56% of which were verified on video [20]. The mean PLA for a definite direct head impact was 47.2 g (n = 37, range 30 to 102 g).
- In 210 male and female professional AFL players, there were 336 HAEs with PLA $\geq$ 30 g. The majority were distributed between 30 and 60 g, but there were a small number of impacts greater than 100 g [21].

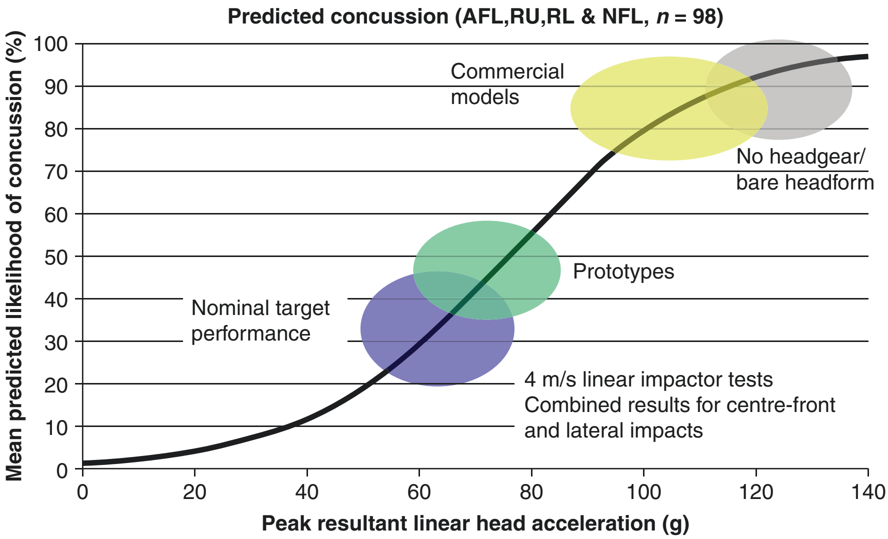

Fig. 5.1 Bare HIII headform and headgear performance in linear impactor tests superimposed onto an injury risk curve. PLA from impact tests and PLA-based concussion injury risk curve formed by pooling NFL and Australian (unhelmeted) football reconstructed injury and non-injury concussion cases

- These data indicate a role for headgear in reducing the severity of the less frequent direct head impacts that are associated with a concussion risk, for example, greater than 75 g on human heads, and managing the more common low severity impacts.

The Australian Football League (AFL) is working towards implementing performance standards for headgear. In short, the basic standard specifies drop tests at four sites, with three repeats, from 300 mm with PLA ≤ 150 g for the first impact and PLA ≤ 200 g on repeat impacts. The advanced specifies drop tests at four sites, with three repeats, from 300 mm with PLA ≤ 100 g for the first impact and PLA ≤ 140 g on repeat impacts and linear impactor tests as described above (4 m/s linear impactor tests with 4 kg mass) with PLA ≤ 75 g and PAA ≤ 7500 rad/s^2. Laboratory testing of prototype designs has demonstrated that these are achievable objectives. Ideally, once a model becomes available that is accepted by players, its effectiveness will be evaluated in a randomized controlled trial. The performance criteria reflect what is achievable currently and other factors, for example, the differences between the dynamic responses of the human head and a rigid headform, that is used in drop tests.

Combat sports A range of headgears intended for use in combat sports were evaluated using drop tests and linear impactor tests [22, 23]. The headgear models were selected because of their characteristics, that is, head coverage, density and thickness. Drop tests were performed with a rigid “M” headform (5.6 kg drop assembly) from 0.2, 0.4, 0.5 and 0.8 m with repeat tests at each site. Linear impactor tests were conducted at 4.11, 6.85 and 8.34 m/s with a Top Ten branded headgear designated for boxing; a glove/fist interface was used.

Some highlights of the drop tests were as follows:

- At 0.5 m drop height the lowest PLA was measured with the Macho Warrior headgear and the greatest was with the Adidas Taekwondo (TKW), 63 g and 546 g, respectively, for the mean of five repeat tests.
- Headgear “bottomed out” typically between 0.5 and 0.8 m drop heights; Macho Warrior would have bottomed out at a drop height greater than 0.8 m and Adidas TKW bottomed out between 0.2 and 0.3 m drop heights.
- There was a progressive reduction in impact performance at each drop height, even when the impact was well within the capacity of the material to attenuate energy.

The drop tests identified the expected differences based on material density and thickness. We wrote [22]:

> “The best performing headguards were either the heaviest—the Rival RHG 10 at 0.53 kg (average thickness 25 mm, density 140 kg/m^3)—or the thickest—the Macho Warrior at 37 mm (mass 0.3 kg, density 130 kg/m^3). The worst performing headguard was the Adidas Taekwondo model, which was the lightest and thinnest headguard. The two Macho brand headguard models had similar foam densities (125 kg/m^3), but the Warrior’s average thickness was 37 mm compared with the Dyna’s average thickness of 25 mm. The additional thickness explained the Warrior’s superior performance. Comparatively, the Macho Warrior was between seven and eight times more effective in reducing headform acceleration compared with the Adidas Taekwondo model in rigid impacts, but with only a difference in mass of 0.09kg. The opportunities available to designers are to (1) maintain the thickness of the headguard and increase its density, (2) increase the thickness and maintain density or (3) do both”.

The liner impactor results indicated that in simulated punches with speeds between 5 and 9 m/s, AIBA-approved boxing headgear, in combination with a glove, offers a large level of protection to the boxer’s head. For example, in 6.85 m/s tests:

- PLA was greatly reduced from 86 and 89 g to 46 and 60 with headgear, respectively, means for lateral and centre front impacts.
- PAA was greatly reduced from 5200 and 5600 rad/s^2 to 2800 and 2900 rad/s^2 with headgear, respectively, means for lateral and centre front impacts.
- Under these punch loads, PLA was greater than a nominal concussion threshold of 75 g without headgear and reduced to less than the threshold with headgear; and, PAA was close to a nominal concussion threshold of 6000 rad/s^2 without headgear and halved with headgear.

In total, the testing of headgear for combat sports showed that the better performing models would offer protection during training and competition. Often, a false dichotomy is discussed regarding headgear, that is, the use of headgear results in poor defensive technique. There is no barrier to training with and without headgear to focus on technique and developing athletes with good technique and who also wear headgear. In motorsports, the pilots and riders adopt the best techniques and equipment.

Projectile sports (Cricket/Baseball) Helmets in cricket and baseball are intended to prevent head injury and provide a structure for mounting a faceguard or visor. The faceguard prevents facial and ocular injury, as well as other head injuries. Despite the similar hazards in the two sports, cricket helmets tend to have a thin relatively stiff liner in contrast to thick and compliant baseball liner. The success of helmets in managing the head impact acceleration in projectile impacts was assessed in a selection of helmets [14]. Standards for cricket helmets have developed in the intervening period and include a projectile test for the faceguard and neck protectors. Our work indicated little correlation between the magnitude of headform accelerations in equivalent impact energy tests conducted using drop tests onto a rigid anvil (as per the current standard) and projectile tests for cricket helmets. In contrast, there was a better correlation between projectile test results and drop tests onto a modular elastomeric programmer anvil for baseball and ice hockey helmets. This demonstrates that impact tests can be developed that do not necessarily resemble sports-specific impact characteristics but are indicative of helmet performance in sports-specific impacts. At that time, baseball helmets demonstrated a greater capacity to reduce headform acceleration than cricket helmets, although the results did not indicate that a baseball or cricket helmet would prevent concussion if the projectile struck the head in an impact directed radially (or centric) to the head's centre of gravity (Table 5.2). However, it is more common in match situations to observe a glancing ball-to-helmet impact.

Cycling Bicycle riding is a major sporting, recreational activity and means of transport. The hazards and injury risks in bicycle riding are broad and large. A cyclist may fall off while cycling and hit the road surface or in a more severe crash may collide with a moving car. As per American football, the initial rationale for bicycle helmets

Table 5.2 Cricket and baseball helmet projectile impact results

	Bare Hybrid III headform		HIII headform with helmet		Per cent reduction relative to bare headform (%)	
Ball speed (m/s)	Cricket PLA (g)	Baseball PLA (g)	Cricket PLA (g)	Baseball PLA (g)	Cricket	Baseball
19	278	316	67	72	76	77
27	347	426	160	139	54	67

Average of the maximum headform acceleration (PLA) is presented for all impact sites combined for bare headform and helmeted impacts with the appropriate ball

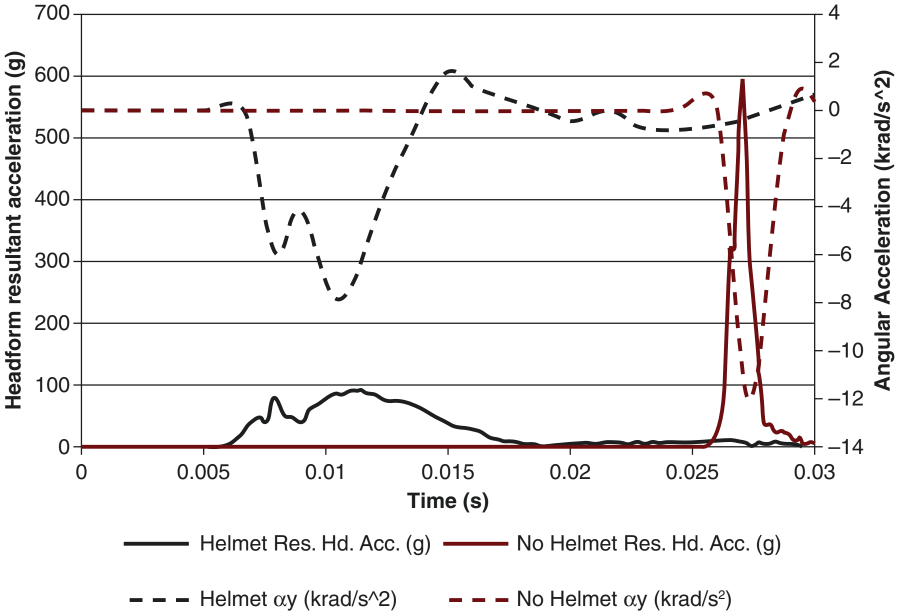

Fig. 5.2 Comparison of head linear and angular acceleration time histories in oblique impacts using a Hybrid III head and neck. Occipital impacts were conducted with a drop height of 1 m and striker (horizontal) speed of 15 km/h. The resultant headform acceleration was around 100 g for the bicycle helmet impact compared to 600 g for the bare headform impact. Peak angular acceleration in the helmet impact was almost half the bare headform impact

was to prevent a more severe spectrum of injury, including skull fracture, intracranial haemorrhage and penetrating wounds, rather than sports concussion. Recent comparative crash simulation tests have demonstrated that the laboratory performance of bicycle helmets is a reasonable predictor of the real-world performance [15]. In comparison to helmeted impacts across all impact configurations, mean maximum headform acceleration was 2.8–6.7 times greater without a helmet and angular accelerations were between 2.0 and 7.3 times greater without a helmet, depending on the exact impact characteristics (Fig. 5.2). An analysis of the oblique test results using biomechanical injury likelihood relationships again paralleled well the results of epidemiological studies. The analyses showed a significant effect of helmets on reducing the likelihood of severe head injury, but a potential for concussion to occur across a range of impacts. In contrast, the bare headform tests predicted a high risk of severe skull and brain injuries even in the more benign crash scenarios.

Heading in football/soccer: why current helmets are not needed Helmets are available and marketed for soccer. There are no convincing epidemiological or laboratory studies that demonstrate their effectiveness or efficacy. Although there is a risk of concussion in soccer, it is relatively low, compared to American football and/

or rugby. We measured PLA in a range of soccer skills from a shoulder collision to a finishing header [24]. For a range of heading drill events, we observed a mean PLA = 15.6 g (SD = 11.8 g) and in a limited number of training situations mean PLA = 20.7 g (SD = 10.6 g). These impacts are substantially lower in severity than in Australian Football. Despite concerns that heading itself may cause brain injury through a cumulative dose effect, the evidence suggests that during the aerial contest for the ball, head impacts causing immediate injury occur because of head-to-head impacts or arm-to-head impacts [25]. These intentional or accidental impacts can be controlled through the laws of the game. Arguably, helmets would reduce the ability of a player to head a ball and may lead to players compensating for the loss of ball rebound by changing their head-neck dynamics. This in turn might result in higher speed head-to-head impacts, although this is speculative. Unlike contact football where accidental head contact does occur frequently, soccer has other opportunities to prevent concussion through its laws, law enforcement, training and supervision. Considering a cumulative head acceleration dose, it is noteworthy that a dose component representing headers would be overwhelmed by PLA induced through non-contact general skills, for example, kicking a ball and re-directional running. The frequency of the non-contact general skills would be an order of magnitude greater than heading and the PLA magnitude associated with heading is not substantially greater than non-contact general skills. This could result in a very active player who experiences very few direct head contacts being considered "at risk" of developing a brain condition as a result of "sub-concussive impacts", when in fact there is no risk and the player's health, fitness and personal satisfaction are potentially compromised by their match and training exposure being reduced as a result of a falsely assessed risk.

Future Development

There is a need for general and sports-specific research and development to improve the protection offered by current helmets. Our understanding of the mechanisms of concussion generally and in specific sports, as well as human tolerance levels, continues to improve. Knowledge in these areas is consistent with established injury criteria for more severe head injuries. When this knowledge is applied to helmet test methods, standards and helmet design improvements in the ability of helmets to prevent concussion can be expected.

Correlations between biomechanical test data for helmets and epidemiological studies are generally high. The trends in improved impact energy attenuation are paralleled between the lab and field studies and absolute measures of head acceleration can predict on field helmet performance, albeit imperfectly. The strengths of the correlations are affected by intrinsic and extrinsic factors and the nature of the inciting event that influence injury likelihood and injury severity on field. These are not necessarily considered fully in laboratory test methods.

Current tolerance data treat concussion as one single pathology although the clinical symptoms and variation in cognitive and other impairments suggest differences within the umbrella term of concussion. Age-specific tolerance data are not available, for example, on children. It is also becoming clearer that impact direction and location influence concussion tolerance. In this context, the use of resultant head linear or angular acceleration criteria may not be optimal. Therefore, test methods will need to develop further.

The role of angular acceleration in concussion is gradually being resolved. It is rare for high angular acceleration to occur without high linear acceleration or impact force. Therefore, these characteristics are typically coupled. Despite the focus of helmet testing on linear acceleration management, helmets do reduce angular acceleration. Further improvements in this area are possible but require suitable test methods and standards, without compromising linear acceleration performance.

If a causal relationship between cumulative head impact exposure and brain injury is conclusively proven, that is, so-called sub-concussive impacts, then helmets in those sports that permit intentional head impact or have a high incidence of accidental head impact will need to offer even greater protection in comparison to protection against a single overload event. At present the objective should be to prevent concussion, because it is a known risk and there are known consequences of repeat concussions.

It is imperative that biomechanical laboratory studies and well-designed epidemiological and neuroimaging studies are conducted together. In comparison to epidemiological studies, laboratory studies are inexpensive and variations can be made and assessed rapidly. Confidence in laboratory studies that can be gained through validation through epidemiological studies assists in a cycle of improvement. Video analysis of games coupled with on-field monitoring of head impact biomechanics, behavioural surveys and usability studies further enhance knowledge gained from epidemiological studies, as these assist in the interpretation of the main epidemiological results.

As a final note, there has been an enormous expansion of biomechanical knowledge in the field of concussion and helmets in sport over the last 20 years. As research findings are translated into helmet design and as new helmet technologies develop, improvements in the ability of helmets to prevent concussion can be expected. This requires the support of the major sports, equipment manufacturers, research groups, public funding bodies, standards organizations and athletes.

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