Experimental investigation into suitability of smart polymers as an impact-absorbing material for an improved rugby headgear

Mladenko Kajtaz, Joseph Karren, and Aleksandar Subic

Abstract**—**Concussion in Rugby is an important public health issue and there is a clear need for concussion-reducing headgear. It is not clear whether currently available soft headgear can mitigate the effects of a concussive impact and it is concerning that a notable proportion of the playing population have a misbelief in their protective effect. In order to enhance development of modern and effective headgear, this study investigated suitability of smart polymer materials via both standardised and novel biomechanical test methodologies. Superior performances of the smart polymers to that of the existing products were generally observed and resulted in the reductions of at least 40% in the probability of concussion. Diffuse Axonal Injury (DAI) severity rating suggested two-fold improvement over the existing headgear if the smart polymers were used as a replacement for the cushioning material in the current products.

Keywords**—** biomechanical head impact testing, concussion, smart polymers, sport

I. Introduction

COMPETITIVE sports offer the ability to participate in a wide variety of school, collegiate, recreational and professional arenas. However, risk of injury is inherent in any sport, especially in contact sports such as Rugby (Rugby Union). Rugby is a ball carrying, collision team sport played at school, collegiate, club and professional levels in more than 100 countries across 5 continents [1]. Contesting for possession of the ball, players must cohesively and forcefully defend their goal line by tackling the ball carrier to interrupt the opposing play. Although tackles above the shoulders of a player are not permitted under the rules of the game, inadvertent head impacts are a common occurrence. As with other types of injury, the frequency and severity of head injury in sport is a function of the nature of the game, the rules, the regulations and the actions of participants. The retirement of several high-profile athletes as a result of concussion has increased the awareness of the sporting community to the pertinence of this issue. Much is unknown about the pathogenesis and long-term implications of sub-concussive and concussive blows, but what is known is that concussion is a complex phenomenon resulting from both linear and rotational injury mechanisms. Even though Rugby headgear are permitted under the rules of the game, their effectiveness in mitigating concussive impacts is not fully understood and their role is contentious. Many advocate that it only protects the wearer's head from abrasions, while others suggest that its role should be to act as a shock absorber. Pettersen [2] investigated the attitudes of Canadian Rugby players and found that most (62%) believe that headgear can protect against concussion. Finch et al. [3] also found that a large proportion of Australian school-age Rugby players believe that wearing soft headgear gives them more confidence in a tackle. A misbelief in the protective ability of headgear may be a catalyst for more severe head injury, so it is crucial for the role of soft headgear to be more adequately defined. As kinetic energy is applied to the cranium, the resulting inertial acceleration/deceleration and rotation of the brain causes internal strains that disrupt the cerebral anatomy and neurological framework [4]. For a protective headgear to function effectively, it must attenuate this impact energy to decrease the magnitude of the impact force to the head [5]. Kemp et al. [6] observed 13 English Rugby premiership teams and found no significant reduction in injury risk among headgear wearers. A randomised cluster controlled trial (RCT) by McIntosh and McCrory [5] involving Australian players yielded similar findings. Laboratory tests [7-9] have also found that many commercially available soft headgear perform poorly under simulated impact conditions.

To make headgear more effective, several studies have suggested the use of thicker, denser padding [8-10], but sanctions within the sport prevent such improvements from being made. The International Rugby Board (IRB) is the governing body for the global sport of Rugby Union, and imposes restrictions on the design and performance of approved headgear (in contrast, Rugby

M. Kajtaz is with RMIT University, School of Aerospace, Mechanical and Manufacturing Engineering, GPO Box 2476, Melbourne, VIC, 3001, Australia (e-mail: mladenko.kajtaz@rmit.edu.au).

J. Karren was with The University of Sydney, School of AMME. He is now with Human Impact Engineering, Sydney, NSW, Australia (e-mail: jkarren@humanimpacteng.com).

A. Subic is with RMIT University, School of Aerospace, Mechanical and Manufacturing Engineering, GPO Box 2476, Melbourne, VIC, 3001, Australia (e-mail: aleksandar.subic@rmit.edu.au).

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League and Australian Rules football do not impose such regulations). The controversy over whether headgear prevent concussion has caused confusion as to their role in the game, and with rising participation and popularity in the sport, more focused research is necessary. Despite these sanctions and restrictions, a few studies have investigated changes to padding thickness and density in new headgear designs. In a laboratory study by Hrysomallis [8] it was suggested that padding of at least 15mm on the front and sides had the potential to offer adequate impact protection. In another investigation, a thickness increase of 5mm significantly reduced the mean linear headform acceleration maxima, concluding that it could be possible to use thicker foams to reduce concussion [9].

Unlike the existing investigations, this paper explores an application of smart polymers in new headgear designs. In addition to the standardized tests, this study in a unique way uses crash test dummies as surrogates to represent Rugby players in a tackle. This utilization of the crash test dummies is embodied in a novel and recently developed sport-specific testing methodology that allows for an introduction of other advanced and more relevant head injury criteria that were not typically a part of the existing and commonly used standardised tests.

II. Head Impact, Concussion and Injury Criterion

A head impact causes both contact and inertial loading at the impact location, whereby the direct focal contact forces at the impact site are linked to skull fracture and brain contusion resulting from skull deformation; while the inertial movement of the brain within the cranial cavity is linked to ‘coup/contre-coup’ type brain contusions (a coup type brain contusion occurs when the rapidly decelerating head forces the cerebrum towards the point of impact, whereas a contre-coup type contusion occurs when the movement of the brain causes negative suction pressures on the side of the brain diametrically opposite to the impact area). During this motion, the brain may also glide over the irregular, jagged contours of the skull causing cerebral lacerations [11]. Mild traumatic brain injury (mTBI), or commonly concussion, is a common head injury associated with many of these head impacts. New understanding of the pathophysiological underpinnings of this injury as well as its short and long-term effects has recently been a cause for concern in the realm of sports medicine [12]. It has been shown that repetitive concussive and subconcussive blows may increase the risk of developing neurocognitive impairment in later life [13]. A large body of the research has shown that concussive injury is primarily the result of the inertial and impulsive loading experienced throughout the brain during impact, rather than the result of acute coup/contre-coup type contusions and skull fractures.

Although, with the head and neck motions that occur in a typical oblique head impact, both linear and rotational forces inside the skull cause intracranial damage, early efforts to understand the biomechanical basis of concussion focused predominantly on linear force/acceleration and its effect on human injury. Lissner et al. [14] attributed intracranial damage to the skull deformation and pressure gradients caused by direct head impacts to cadaver heads and developed the Wayne State Tolerance Curve (WSTC) (see **Fig 1**), implying that linear acceleration has a strong association with brain injury pathogenesis and it could be adapted and used as a criterion for milder brain injury types [15, 16]. Today these indices are used to measure the likelihood of head injury in a range of vehicle, personal protective gear, and sports equipment standards for the prevention of both mild and severe forms of TBI. However, because they are based purely on linear acceleration, their validity in assessing concussion has been debated and many preeminent researchers have questioned their ability to diagnose injury risk because most impacts also involve rotational loading on the head. Importantly, brain tissue deforms readily in response to rotational forces since the bulk modulus of brain tissue is approximately 10^5 times its shear modulus, hence there is a general consensus in the literature that shear deformation caused by rotational motion is the predominant mechanism of brain injury [17, 18]. Studies showed [19] that it is difficult to produce traumatic unconsciousness in the absence of rotational motion but the likelihood of an unconscious episode is substantially increased once it is introduced. Early research by Ommaya and Hirsch [20] estimated that angular acceleration greater than 1,800 rad/sec^2 is likely to cause cerebral concussion in humans; while. Pincemaille et al. [21] established a concussion threshold of 13,000-16,000 rad/sec^2. Currently, rotational criteria are not incorporated into a vast majority of sports and recreational helmet standards due to the lack of extensive evaluation and review needed for them to be accepted by the scientific and sporting community.

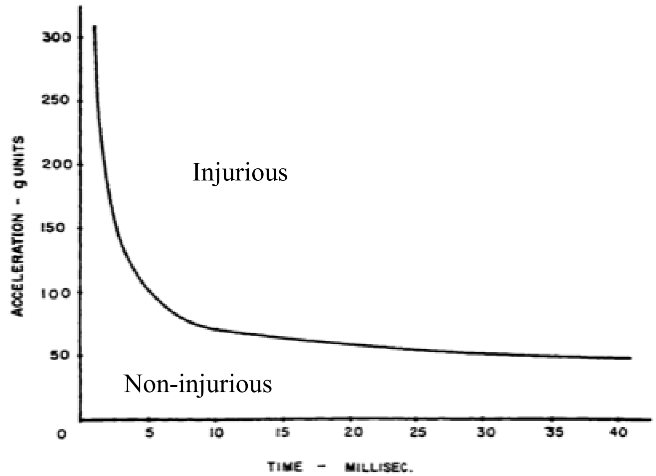

Fig 1. Wayne State Tolerance Curve, plotting acceleration as a function of impact duration to predict injury

However, recent investigations have made use of new technology to more accurately measure human tolerance to both linear and rotational acceleration in sport. In particular, many injury biomechanics researchers have investigated mild brain injury using accelerometers placed inside American football players' helmets. By observing the acceleration data of medically validated cases of concussion, these studies have proposed thresholds ranging from 70 to 165 g in padded impacts [22-25]; or a linear injury threshold of 78 g and a rotational threshold of 6322 rad/s^2 to represent a 50% risk of concussion, as proposed by Newman et al. [26]. Similarly, Rowson et al. [27] analysed concussions in a large number of NFL helmet impacts and proposed a linear acceleration injury threshold of 80g and a rotational threshold of 6383 rad/s^2 and 28.3 rad/s (peak resultant angular acceleration and maximum change in resultant angular velocity respectively). Other studies have observed similar thresholds for concussion [23, 24, 28, 29].

III. Existing Rugby Headgear

As head impact involves a transfer of energy and momentum due to changes in the relative velocities between the head and another object, a concept of placing a protective material between the head and the striking object to attenuate and distribute the impact force over a larger area is the basis for wearing head protection. Protection, as a safety concept, is defined as a reduction of impact severity to a level that the brain can tolerate whereby the tolerance is defined by an injury threshold/criterion. Since the majority of sports and recreational helmet standards specify the safety criteria in terms of linear acceleration only, current helmets are primarily designed to reduce the linear acceleration component of head impact [30].

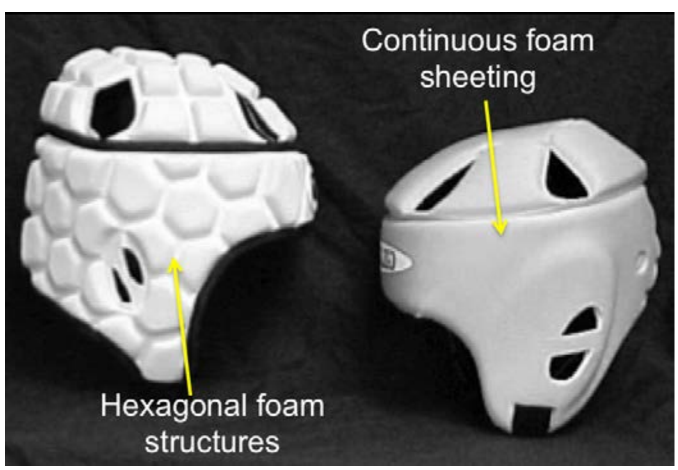

Fig 2. Typical headgear construction – honeycomb (left) and continuous (right).

There are several types of protective headgear available for use in Rugby Union, which typically have thickest padding at the forehead and sides, and thinnest at the top and back of the head. Honeycomb configurations (**Fig 2a**), featuring small, polygonal-shaped blocks of foam under a fabric skin, and continuous configurations (**Fig 2b**), constructed of larger foam sheets that extend over a greater area; are two most common designs of headgear construction. Impact testing by McIntosh et al. [9] found that the

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continuous arrangement performed better than the honeycomb arrangement due to improved load distribution and lateral stability, however it seems that the aesthetic and/or comfort value honeycomb-type headgear has resulted in their popularity in recent years and a decline in the availability of continuous types.

A. Use of protective headgear in Rugby

Pettersen [2] investigated headgear-wearing rates in Canadian Rugby players, and found that while 62% believed that headgear could offer them protection from head injury, only 26% admitted to wearing them. Primary reasons for not wearing headgear were that it caused players to become “too hot”, “uncomfortable” and impeded their ability to communicate on field. An Australian study by [3] found that over 70% of school-age Rugby players reported to regularly wear soft headgear and 67% believed that wearing headgear made them feel “safer” on field. Both these studies observed an assumption that headgear is protective. Garraway et al. [31] suggested that players use protective headgear with the expectation that it will reduce the severity of impact so they can tackle harder when wearing them. Such “risk-compensation” by players may result in a paradoxical increase in the risk of head injury. It is especially concerning that most players who wear headgear are young players who, as mentioned above, exhibit higher vulnerability to concussion compared to mature players. However, the idea that athletes wearing headgear will play more aggressively has yet to be substantiated, and despite the predominant belief that headgear are protective, it has been shown through the use of an instrumented tackle bag that players do not tackle harder when wearing headgear [32].

B. Performance and Safety Aspects of Existing Rugby Helmets

The presumption that headgear protects the wearer from concussion makes their role in Rugby very controversial. While some argue that they should be used for protection against impact, others believe that they should only be used to reduce lacerations and abrasions to the scalp. Prevention of lacerations and abrasions has been confirmed by a number of studies, including an investigation by [33] who concluded that headgear were associated with a significant reduction in bleeding head injuries in a cohort of English Rugby players. On the other hand, a number of investigations on the impact attenuation of headgear have revealed their poor performance and there is currently not enough evidence to suggest that they can protect the wearer from concussion [6, 34, 35]. Similarly, various laboratory studies have also found headgear do not attenuate sufficient impact energy to be able to protect players on the field [7, 36, 37].

IV. Rugby Headgear Testing

In order to evaluate the protective capability of soft headgear without human subjects, head impact attenuation tests are devised to reconstruct concussive tackles. These sport-specific test methodologies need to simulate typical head impacts so that the results of these tests are delineated to the playing field. Therefore, helmet/headgear performance test methods tend to contain three essential items: a test system that is a good representation of reality; a durable headform that has a similar dynamic and kinematic response as the human head; and an injury criterion that is applicable to the test configuration. Van den Bosch [38] developed a load-injury head impact model (see **Fig 3**), which highlights the predicament faced when attempting to replicate head impacts in a laboratory setting. In particular, during a Rugby head impact, a mechanical load is applied to the head which will cause a unique biomechanical response that depends on: the impact attenuation characteristics of the headgear, the interaction between the head and the headgear and the head impact site (e.g. side, front, back). A tissue damage by means of linear and rotational forces is assumed if the biomechanical response exceeds a certain threshold for a specific impact type. Obviously, this head impact testing offers a means to crudely simulate the loading conditions and dynamic responses of a human collision. Even then, determining the injury risk depends on injury assessment tools that are largely unrefined, as mentioned previously.

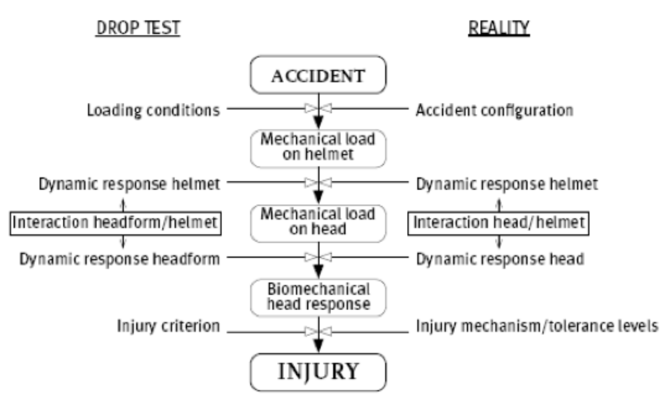

Fig 3. Load-injury head impact model comparing a laboratory test to reality [38]

A. Helmet Testing Standards

A typical headgear performance testing involves similar methods to the AS/NZS 2512.1.1 head drop system (see **Fig 4**), which uses a magnesium alloy artificial headform guided in free-fall from a height onto a rigid, flat surface. Accelerometers in the headform measure its response and any effect of headgear in dissipating the impact energy. The majority of the abovementioned literature regarding headgear effectiveness (e.g. [5, 9], and the IRB standard [39], have utilised this form of drop test. Historically, this methodology has not been entirely satisfactory in evaluating headgear performance as it fails to represent the actual character of brain trauma [40]. Furthermore, this headgear evaluation test has a number of distinct limitations. Firstly, the use of a linear head drop system only assesses the linear acceleration of the impact and does not account for rotational injury mechanisms. Second, it also does not reflect the speed and severity of collisions experienced by Rugby players. The drop height of 300 mm specified in the IRB performance standard delivers impacts (2.4 m/s, 14.7 J) that are of significantly lower severity than the collisions experienced by Rugby players (5±1 m/s, 50-60 J). Third, in these tests there is no requirement for headform stiffness to correlate with that of the human skull, nor is there a requirement for the impacting surface to match any object likely to impact the player on field. Rigid metal headforms have demonstrated unrealistic impact responses compared to cadaver models and there is no way to delineate the human equivalent of a rigid headform impact, nor the protective capacity of any helmet using this technique (Mills 1990). Hrysomallis [8] argued that since bodily segments respond with motion and deformation during a head impact, deformable surfaces should be used to simulate real play. A possible explanation for incorporating hard surfaces may be in inability to produce repeatable results if the surfaces were allowed to deform [40].

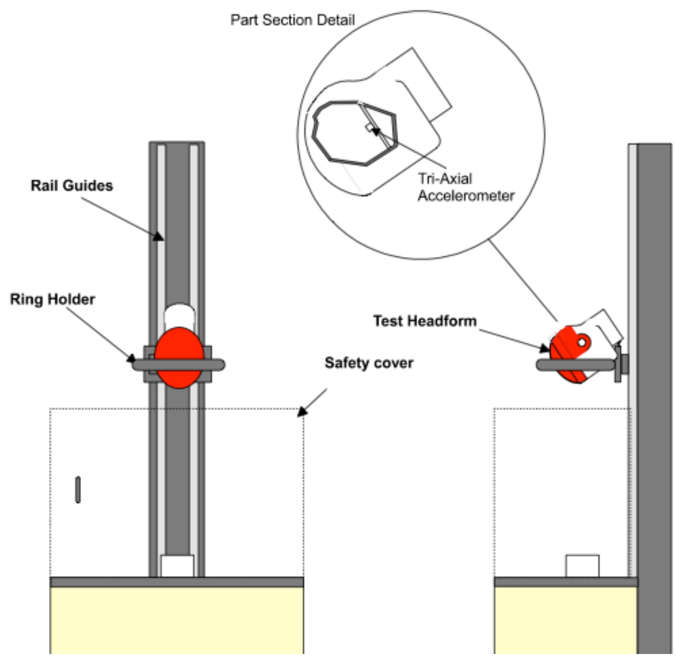

Fig 4. Standardised head drop system used by the IRB and others, similar to AS/NZS 2512. [39].

V. METHODOLOGY

The impact attenuation potential of alternative and non-traditional materials (i.e the smart polymers) was investigated using two laboratory tests. The first test methodology was the standardised head drop system used by the IRB and others, which was extensively discussed in the previous sections. Despite its numerous distinct limitations, this test is the only standardised and widely recognised methodology, which provides a basis for all further comparisons. The second test methodology was a novel, non-standard and sport-specific test methodology developed to reconstruct a typical concussive head impact scenario in Rugby

A. The Standardised Test Methodology

A guided drop test rig was used to impact a 5.5 kg hemispherical impactor (D=145mm) instrumented with a uniaxial accelerometer onto stationary flat sheets of the sample impact-absorbing materials (see **Fig 5**). The history of acceleration was recorded and the peak acceleration was presented as g-force as per the IRB Regulation 12 [39]. The samples were all tested at the same ambient temperature of 23C.

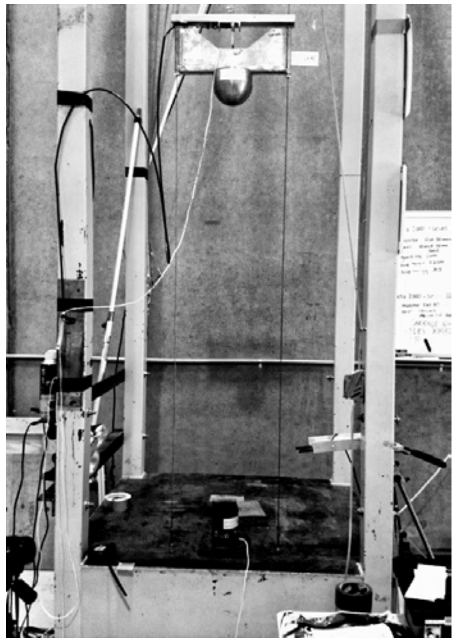

Fig 5. A setup of the standard impact drop test

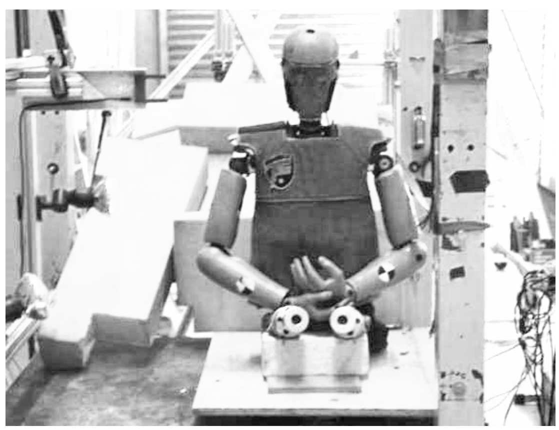

Fig 6. THOR - an advanced impact dummy

B. The New Sport-Specific Test Methodology

This test methodology was developed by one of the authors [41, 42] and consists of utilising a THOR (Test Device for Human Occupant Restraint) ATD to reconstruct a typical concussive head impact scenario in Rugby. Shown in **Fig 6**, THOR is an advanced impact dummy and is based on improved biomechanical knowledge compared to the commonly used Hybrid III, making it one of the most biofidelic devices available [43]. THOR also uniquely incorporates a specifically designed shoulder complex that simulates human shoulder seatbelt loading in frontal crashes [44, 45] that allowed this dummy to be used in a controlled impact-testing environment to reconstruct concussive shoulder-head tackles.

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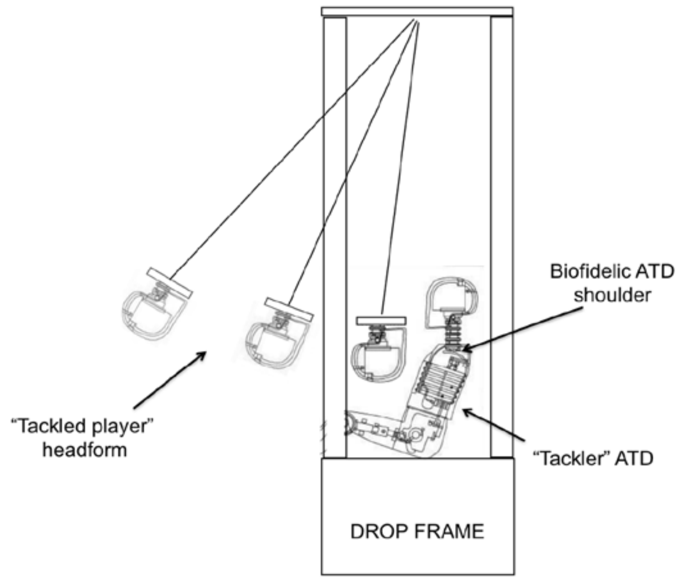

Fig 7. Schematics of the new biomechanical and rugby-specific testing setup

Fig 7 schematically shows the test setup by depicting a pendulum drop testing apparatus that guides a swinging THOR headform into impact with the acromioclavicular region (shoulder) of a THOR torso, similarly to the head of a tackled player coming into contact with the tackler's shoulder. Instrumentation within the headform measured the linear and rotational acceleration at the centre of gravity (cg) of the headform as it collided with the surrogate shoulder. An impact velocity of 4.65 m/s, which is in the range of impact velocities found to cause concussion in Rugby players (5±1 m/s) was used in this test and only temporoparietal impacts (side impacts) were considered in this study (see **Fig 8**).

Fig 8. Temporoparietal "patch" samples modelled off the side region of a standard headgear.

C. Material Test Samples – Material Solutions

Test samples of the impact-absorbing materials were prepared from flat material sheets provided by commercial suppliers, namely Rogers Corporation from the USA, D3O from the UK and Albion Sports from Australia.

Poron XRD is a lightweight, thin and breathable material that's engineered for repeated impact and shock absorption, provided by Rogers Corporation. Poron XRD is a urethane based and highly damped strain-rate sensitive material with a memory-foam-like properties that exploits the glass transition temperature (Tg) of the urethane molecules to enhance its impact attenuation properties. Three different densities of Poron XRD were investigated in this study (144, 192 and 240 kg/m3). The thickness for all was 9.6mm.

D3O is a British-based impact protection solutions company that markets a unique patented technology used to produce a shock absorbing material. Using a proprietary formula to engineer non-newtonian materials, D3O has the ability to engineer a number of dilatant materials and, using a patented technology, incorporate them into polymer. Two of their materials AERO (220 kg/m3) and DECELL (330 kg/m3), with two different thicknesses (6 and 10mm) were investigated in this study. The AERO material is a urethane based like Poron XRD, while the DECELL is based on an ethylene vinyl acetate (EVA).

Traditional foams were provided by Albion Sports, the Australian leading manufacturer in sports protection equipment and a

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former manufacturer of Rugby headgear. Samples of their EVA (40 kg/m3) and expanded polystyrene (EPS) (80 kg/m3) protective liners that are used in their hard shell helmets (cricket and cycling helmets) were included in this comparative study as examples of impact-absorbing materials used in other sports helmets. The test thickness for both materials was 10mm.

Headgear: Samples of two generic models of IRB-approved rugby soft helmets were obtained from a supplier of retail rugby headgear: an entry level; and a top end headgear. The entry level headgear was easily identifiable by a poorer quality of manufacturing upon an inspection.

VI. Results

A. Results of the Standard Tests

A total of 13 different solutions were investigated resulting in 39 individual tests. The results, shown in **Fig 9**, are averages of three repeated trials for each solution and since minimal variations between the trials were observed, the overall repeatability was deemed acceptable. Peak linear acceleration, as the injury assessment criterion in this test, was expressed in terms of "g-force". As expected, worst performing solutions were commercial headgear with their peak linear accelerations in excess of 350g because they were designed to adhere the IRB Standard, which enforces the minimum level of impact attenuation performance (i.e. cannot be lower/better than 200g) and the upper limit was historically kept to 550g. According to LOGIST plot relating the risk of concussion to peak linear accelerations that were developed by [46] (see **Fig 10**), such an inadequate performance of the commercial headgear undermines the philosophy of protective equipment in general, and demonstrate that the IRB does not support the use of headgear for impact protection.

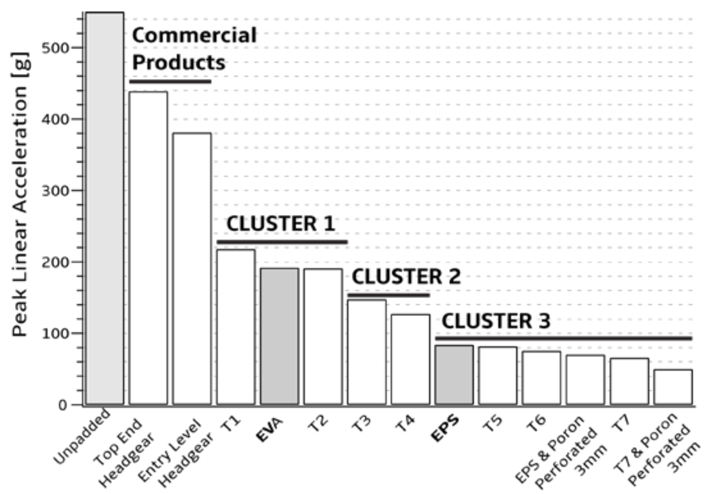
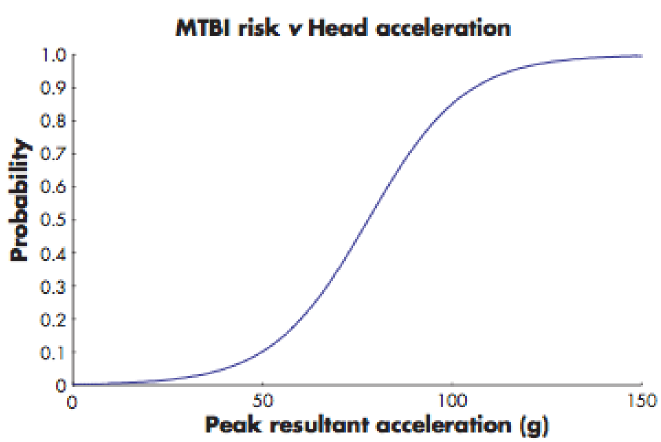


Fig 9. Result of the standard impact testing



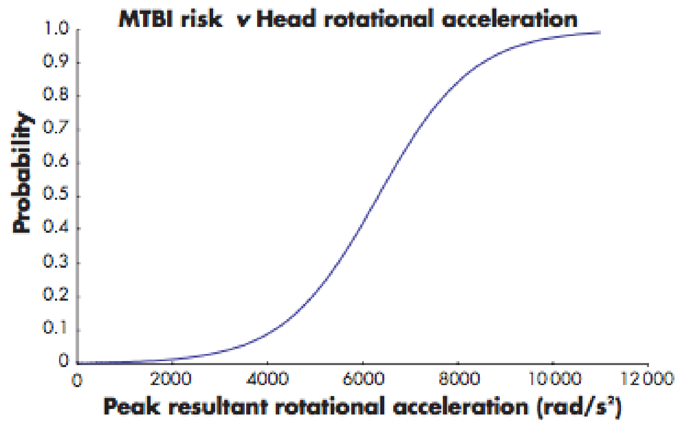


Fig 10. Probability risk curve for linear (top) and rotational accelerations (bottom) [46].

For confidentiality reasons, non-identifiable labelling of the smart polymer materials needed to be used, thus trends and general observations are discussed rather than the individual solutions. As it can be observed, the range of observed peak linear accelerations values is significantly large, which ultimately suggests a possibility of finding a balanced solution that encompasses an increased protection and an elimination of the concerns that such headgear may be a catalyst for more severe head injury. It can also be observed that the solutions could be grouped into three clusters based on their performance. The first cluster ("Cluster 1"), grouped around the EVA solution, includes soft and under-performing solutions and although these solutions may provide significant improvements to the existing rugby headgear, they still fail to reduce any risk of concussion (i.e. >150g). The second cluster ("Cluster 2") can be considered as the solutions with a transitional and relatively mediocre performance with some tendency to reduce the risk of concussion. The EPS solution separates the final cluster ("Cluster 3") and high performing solutions from the rest. These solutions are capable of at least halving the probability of concussion. When multi-layered solutions were considered, an addition of a 3mm thick layer of perforated Poron XRD to the EPS foam, improved its impact attenuation performance by 10%. The best performing solution was a multi-layered solution consisting of a high performance solution and the 3mm thick layer of perforated Poron XRD, which reduced the risk of concussion below 10%.

B. The New Sport-Specific Test Methodology Results

Unlike the standard test methodology which only had the peak linear acceleration as the injury assessment criterion, this newly developed test allowed the head injury risk to be quantified via several parameters: the peak linear acceleration, angular acceleration and angular velocity. Withnall et al. [46] also developed the LOGIST plot representing the injury risk for peak angular acceleration (see **Fig 10**). Furthermore, a biofidelic injury criterion called the Cumulative Strain Damage Measure (CSDM05) available via the simulated injury monitor (SIMon) finite element head model [47], was additional criterion used in this test methodology. The CSDM calculates the cumulative proportion of brain volume that has reached a threshold of ≥5% stretch over the duration of the load from the three-dimensional kinematic data of the headform [47-50]. Only the smart polymer materials and the commercial headgear were tested using this methodology, resulting in 18 individual tests. The results, shown in **Figs 11a-d**, are averages of two repeated trials for each solution and since minimal variations between the trials were observed, the overall repeatability was deemed acceptable. Since both the headform and the surrogate shoulder were deformable, the test reference involving the bare headform impact was also produced. As it can be observed from the figure, the best performing foam (T1) gave an approximate 37% reduction on the bare headform Peak g while the worst performing foam (T2) gave only around 15% of the reduction. The top three performing solutions reduced the concussion likelihood from 50% to 10% according to the LOGIST curves in **Fig 10**. Rotational acceleration was lower in the across all solutions compared to the bare headform condition, and the top three performing solutions offered reductions of approximately 40% in the probability of concussion. The angular accelerations were similar in order performance to the linear acceleration results for each solution, however angular velocity was relatively constant across all tests. According to SIMon, the lowest and the highest DAI severity ratings (CSDM05: 0.096, 81% reduction on the bare headform performance; and CSDM05: 0.311, 38% reduction, respectively) were assigned to two solutions with ***similar*** linear and rotational performance measures. Similarly, the repeated trials of the same solutions that had very similar linear and rotational components would show high variability in the CSDM05 value, which suggested repeatability issues. Bartsch et al. [51] reported a similar situation in their study and they attribute this inherent variability to high sensitivity of the FE model to rotational acceleration of the head about the coronal plane, which again may emphasizes the importance of considering the rotational acceleration as a vital part of any head injury criterion and targeting rotational acceleration reduction in future headgear may be critically important to their overall effectiveness

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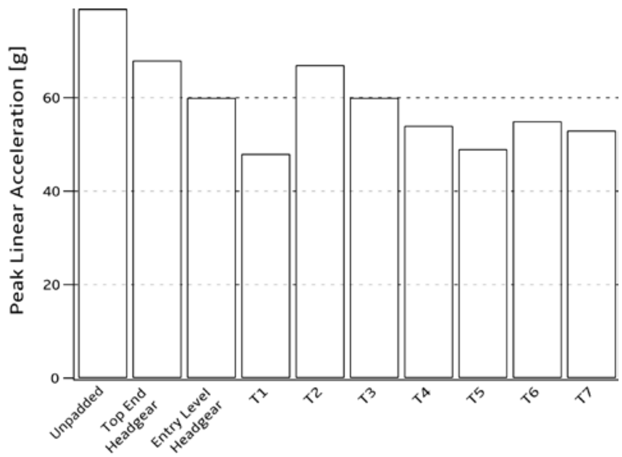

Fig 11a. Results of the side impact foam samples testing, showing linear acceleration

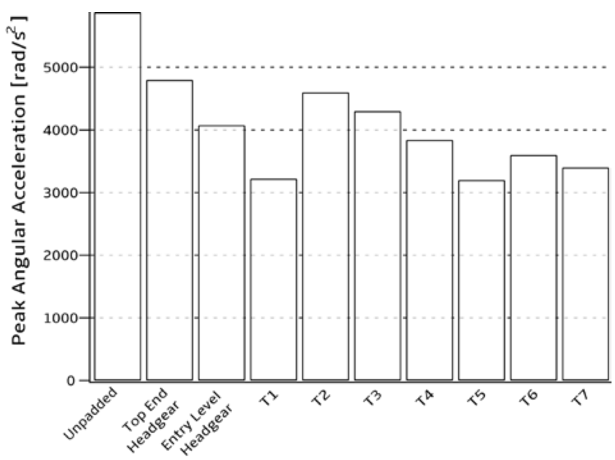

Fig 11b. Results of the side impact foam samples testing, showing angular acceleration

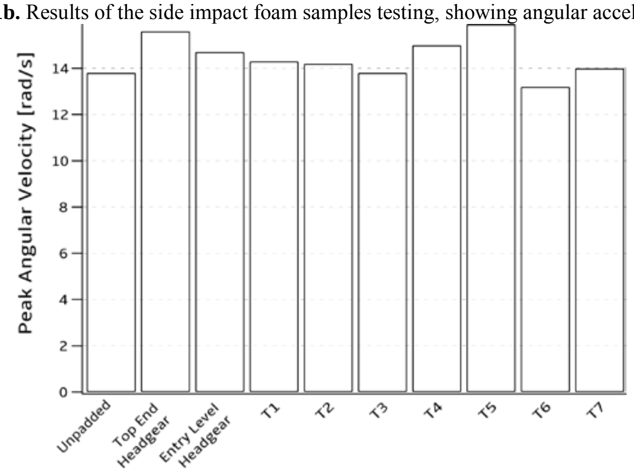

Fig 11c. Results of the side impact foam samples testing, angular velocity

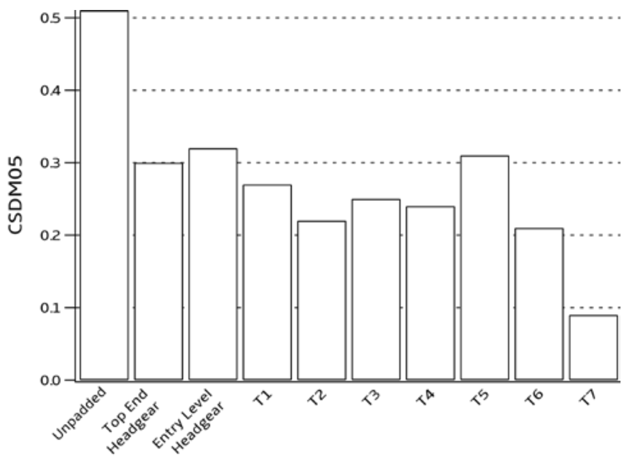

Fig 11d. Results of the side impact foam samples testing, showing CSDM05

C. Performance Comparison across the Two Tests

The peak linear acceleration, as the common parameter in both tests, contrasts the test methodologies and also provides additional insight into the performances of the tested materials (see **Fig 12**). An immediate distinction is the order of magnitudes between the outputted values. While the unpadded contact between an impactor and a base in the standard test was capped to 800g, the bare headform impact in the new biomechanical test was only 80g. This observation is in an agreement with other similar studies [7, 8, 37], which may demonstrate that the incorporation of yielding impact surfaces is an important improvement to the conventional linear head drop system. The trends and the performance of the tested materials generally tend to be corresponding and translate adequately but with one exception (T1) that was confirmed to be valid reading (consistent values during the repeatability testing). In order to suggest an explanation, the mechanical properties of the individual solution would need to be discussed, which carries a risk of identifying the material. However, it is another consequence of the introduction of yielding impact surfaces. Another noticeable distinction is that the erstwhile superior performance of the smart polymers is less augmented in the biomechanical test results, which may also demand a different approach to post-processing of results from a typical comparison of face values (eg. different scaling or performance indicators in order to decompress or increase a differentiation between points).

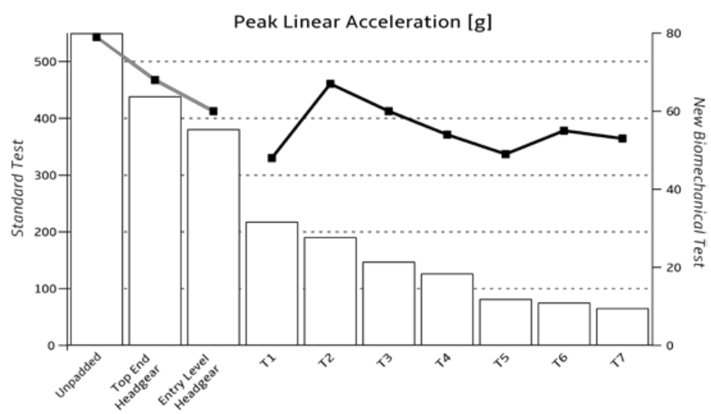

Fig 12. Performance comparison across the tests – peak linear acceleration

VII. Conclusions

The rate and severity of concussion in the sport of Rugby is concerning in light of recent findings relating repeated head impacts to neurological dysfunction and degeneration. Since much is still unknown about causes and effects of concussion, the minimization and the prevention of mild brain injury is of paramount importance to the short and long term health of all who play contact sports. Utilising a recently developed repeatable biomechanical test method to better recreate a common impact scenario resulting in concussion to Rugby players, this study explored application of smart polymers in new headgear designs by comparing their impact attenuation properties with those of the current products and more traditional foams. This study found:

- The current products do not provide sufficient protection.
- *Based on the peak linear acceleration injury criterion:* The majority of the solutions based on the tested smart polymer materials is capable of reducing the probability of concussion below 50%, according to the probability risk curve for

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linear acceleration.

- The best performing solution was a multi-layered solution that reduced the risk of concussion below 10%
- *Based on the peak rotational acceleration injury criterion*: A smart polymer material is capable of offering reductions of approximately 40% in the probability of concussion according to the probability risk curve for rotational acceleration.
- *Based on a biofidelic injury criterion Diffuse Axonal Injury (DAI) severity rating*: A smart polymer material is capable of offering reductions of approximately 80% in contrast to the bare biomechanical headform performance, which is more than twice the reduction of the current products.

REFERENCES

[1] Patricios, J.S. and S. Kemp, "Chronic traumatic encephalopathy: rugby's call for clarity, data and leadership in the concussion debate," *British Journal of Sports Medicine*, vol. 48, no. 2, p. 76-79, 2014.

[2] Pettersen, J.A., "Does rugby headgear prevent concussion? Attitudes of Canadian players and coaches," *British Journal of Sports Medicine*, vol. 36, no. 1, p. 19-22, 2002.

[3] Finch, C.F., A.S. McIntosh, and P. McCrory, "What do under 15 year old schoolboy rugby union players think about protective headgear?," *British Journal of Sports Medicine*, vol. 35, no. 2, p. 89-94, 2001.

[4] Giza, C.C. and D.A. Hovda, "The Neurometabolic Cascade of Concussion," *Journal of Athletic Training*, vol. 36, no. 3, p. 228-235, 2001.

[5] McIntosh, A.S., P. McCrory, and J. Comerford, "The dynamics of concussive head impacts in rugby and Australian rules football," *Medicine & Science in Sports & Exercise*, vol. 32, no. 12, p. 1980-1984, 2000.

[6] Kemp, S.P., et al., "The epidemiology of head injuries in English professional rugby union," *Clinical Journal of Sport Medicine*, vol. 18, no. 3, p. 227-234, 2008.

[7] Knouse, C.L., et al., "Efficacy of Rugby Headgear in Attenuating Repetitive Linear Impact Forces," *Journal of Athletic Training*, vol. 38, no. 4, p. 330-335, 2003.

[8] Hrysomallis, C., "Impact energy attentuation of protective football headgear against a yielding surface," *Journal of science and medicine in sport*, vol. 7, no. 2, p. 156-164, 2004.

[9] McIntosh, A., P. McCrory, and C.F. Finch, "Performance enhanced headgear: a scientific approach to the development of protective headgear," *British Journal of Sports Medicine*, vol. 38, no. 1, p. 46-49, 2004.

[10] McIntosh, A.S., et al., "Rugby headgear study," *Journal of Science and Medicine in Sport*, vol. 6, no. 3, p. 355-358, 2003.

[11] Bailes, J.E. and V. Hudson, "Classification of Sport-Related Head Trauma: A Spectrum of Mild to Severe Injury," *Journal of Athletic Training*, vol. 36, no. 3, p. 236-243, 2001.

[12] Slobounov, S., et al., "Sports-related concussion: ongoing debate," *British Journal of Sports Medicine*, 2013.

[13] Gavett, B.E., R.A. Stern, and A.C. McKee, "Chronic traumatic encephalopathy: a potential late effect of sport-related concussive and subconcussive head trauma," *Clinical Sports Medicine*, vol. 30, no. 1, p. 179-188,xi, 2011.

[14] Lissner, H.R., M. Lebow, and F.G. Evans, "Experimental studies on the relation between acceleration and intracranial pressure changes in man," *Surgery, gynecology & obstetrics*, vol. 111, p. 329-338, 1960.

[15] Gadd, C., "Use of a weighted impulse criterion for estimating injury hazard," *10th Stapp Car Crash Conference Proceedings*, p. 164-174, 1966.

[16] Ono, K., et al., "Human head tolerance to sagittal impact. Reliable estimation deduced from experimental head injury using subhuman primates and human cadaver skulls," *SAE Technical Paper* 801303, 1980.

[17] Unterharnscheidt, F. and L.S. Higgins, "Traumatic lesions of brain and spinal cord due to nondeforming angular acceleration of the head," *Texas reports on biology and medicine*, vol. 27, no. 1, p. 127-166, 1968.

[18] Gennarelli, T.A., et al., "Diffuse axonal injury and traumatic coma in the primate," *Annals of neurology*, vol. 12, no. 6, p. 564-574, 1982.

[19] Ommaya, A.K. and T. Gennarelli, "Cerebral concussion and traumatic unconsciousness correlation of experimental and clinical observations on blunt head injuries," *Brain*, vol. 97, no. 4, p. 633-654, 1974.

[20] Ommaya, A.K. and A.E. Hirsch, "Tolerances for cerebral concussion from head impact and whiplash in primates," *Journal of Biomechanics*, vol. 4, no. 1, p. 13-21, 1971.

[21] Pincemaille, Y., et al., "Some new data related to human tolerance obtained from volunteer boxers," *Proceedings of the 33rd Stapp Car Crash Conference*, 1989.

[22] Funk, J.R., et al., "Biomechanical risk estimates for mild traumatic brain injury," *Annual proceedings/Association for the Advancement of Automotive Medicine*, vol. 51, no., p. 343-361, 2006.

[23] Pellman, E.J., D.C. Viano, and null, "Concussion in professional football: summary of the research conducted by the National Football League's Committee on Mild Traumatic Brain Injury," *Neurosurgical focus*, vol. 21, no. 4, p. E12, 2006.

[24] Pellman, E.J., et al., "Concussion in professional football: helmet testing to assess impact performance – Part 11," *Neurosurgery*, vol. 58, no. 1, p. 78-96, 2006.

[25] Rowson, S., et al., "Linear and angular head acceleration measurements in collegiate football," *Journal of Biomechanical Engineering*, vol. 131, no. 6, p. 61016, 2009.

[26] Newman, J., et al., "A new biomechanical assessment of mild traumatic brain injury. Part 2: results and conclusions," *Proceedings of the 2000 international IRCOBI Conference on the Biomechanics of Impact, Montpellier, France*, 2000.

[27] Rowson, S., et al., "Rotational head kinematics in football impacts: an injury risk function for concussion," *Annals of biomedical engineering*, vol. 40, no. 1, p. Jan-14, 2012.

[28] Zhang, L., K.H. Yang, and A.I. King, "A proposed injury threshold for mild traumatic brain injury," *Journal of Biomechanical Engineering*, vol. 126, no. 2, p. 226-236, 2004.

[29] Duma, S.M., et al., "Analysis of realtime head accelerations in collegiate football players," *Clinical Journal of Sport Medicine*, vol. 15, no. 1, p. 3-9, 2005.

[30] McIntosh, A.S., et al., "Sports helmets now and in the future," *British Journal of Sports Medicine*, vol. 45, no. 16, p. 1258-1265, 2011.

[31] Garraway, W.M., et al., "Impact of professionalism on injuries in rugby union," *British Journal of Sports Medicine*, vol. 34, no. 5, p. 348-351, 2000.

[32] McIntosh, A., C. Caponecchia, and J. Usman, "Tackling risk compensation – a psycho-physical approach to measuring behaviour change," *British Journal of Sports Medicine*, vol. 45, no. 4, p. 362, 2011.

[33] Jones, S.J., et al., "Effectiveness of rugby headgear in preventing soft tissue injuries to the head: a case-control and video cohort study," *British Journal of Sports Medicine*, vol. 38, no. 2, p. 159-162, 2004.

[34] McIntosh, A.S. and P. McCrory, "Effectiveness of headgear in a pilot study of under 15 rugby union football," *British Journal of Sports Medicine*, vol. 35, no. 3, p. 167-169, 2001.
[35] McIntosh, A.S., et al., "Does padded headgear prevent head injury in rugby union football?," *Medicine & Science in Sports & Exercise*, vol. 41, no. 2, p. 306-313, 2009.
[36] Kuncir, E.J., R.W. Wirta, and F.L. Golbranson, "Load-bearing characteristics of polyethylene foam: an examination of structural and compression properties," *Journal of rehabilitation research and development*, vol. 27, no. 3, 1990.
[37] McIntosh, A.S. and P. McCrory, "Impact energy attenuation performance of football headgear," *British Journal of Sports Medicine*, vol. 34, no. 5, p. 337-341, 2000.
[38] Bosch, H.L.A.v.d., "Crash helmet testing and design specifications," *PhD Thesis, Technische Universiteit Eindhoven,* Eindhoven, 2006.
[39] IRB, "Regulation 12: Provisions Relating to Players' Dress," *International Rugby Board*, Dublin, 1998.
[40] Viano, D.C., et al., "Injury biomechanics research: an essential element in the prevention of trauma," *Journal of Biomechanics*, vol. 22, no. 5, p. 403-417, 1989.
[41] Karren, J., et al., "Design of a Test Methodology to Measure Rugby Headgear Effectiveness," *9th Australasian Biomechanics Conference*, Wollongong, Australia, 2014.
[42] Karren, J., "Effectiveness of Rugby Headgear in the Prevention of Concussion," *Honours Thesis, School of Aeronautical, Mechanical and Mechatronic Engineering, University of Sydney*, Sydney, 2014.
[43] Whyte, T., et al., "Dead Injury and Effective Motorcycle Helmets," *Proceedings of Enhanced Safety of Vehicles Conference, Seoul*, Paper Number 13-0108, 2013.
[44] Shaw, G., J. Crandall, and J. Butcher, "Comparative evaluation of the THOR advanced frontal crash test dummy," *International Journal of Crashworthiness*, vol. 7, no. 3, p. 239-254, 2002.
[45] Vezin, P., et al., "Comparison of Hybrid III, Thor-alpha and PMHS Response in Frontal Sled Tests," *Stapp Car Crash Journal*, vol. 46, no., p. 1-28, 2002.
[46] Withnall, C., et al., "Biomechanical investigation of head impacts in football," *British Journal of Sports Medicine*, vol. 39, Suppl 1, p. i49-57, 2005.
[47] Takhounts, E.G., et al., "Investigation of Traumatic Brain Injuries Using the Next Generation of Simulated Injury Monitor (SIMon) Finite Element Head Model," *Stapp Car Crash Journal*, vol. 52, p. 1-33, 2008.
[48] Bandak, F., et al., "Simon: a simulated injury monitor; application to head injury assessment," *SAE Technical Paper*, vol. 2001-06-0222, 2001.
[49] Takhounts, E.G., et al., "On the development of the SIMon finite element head model," *Stapp car crash journal*, vol. 11, no. 47, p. 107-33, 2003.
[50] Marjoux, D., et al., ""Head injury prediction capability of the HIC, HIP, SIMon and ULP criteria"," *Accident Analysis & Prevention*, vol. 40, no. 3, p. 1135-1148, 2008.
[51] Bartsch, A., et al., "Impact test comparisons of 20th and 21st century American football helmets: Laboratory investigation," *Journal of neurosurgery*, vol. 116, no. 1, p. 222-233, 2012.