Review Article

A Review of Instrumented Equipment to Investigate Head Impacts in Sport

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Contact, collision, and combat sports have more head impacts as compared to noncontact sports; therefore, such sports are uniquely suited to the investigation of head impact biomechanics. Recent advances in technology have enabled the development of instrumented equipment, which can estimate the head impact kinematics of human subjects *in vivo*. Literature pertaining to head impact measurement devices was reviewed and usage, in terms of validation and field studies, of such devices was discussed. Over the past decade, instrumented equipment has recorded millions of impacts in the laboratory, on the field, in the ring, and on the ice. Instrumented equipment is not without limitations; however, *in vivo* head impact data is crucial to investigate head injury mechanisms and further the understanding of concussion.

1. Introduction

The potential for concussion is related to the number of opportunities within a sport for events that cause contact to the head of an athlete; therefore, relatively higher incidence rates of concussion are expected in contact, collision, and combat sports compared to noncontact sports. This renders such sports uniquely suited to the investigation of head impact biomechanics. For over half a century, researchers have attempted to use instrumented sporting equipment to measure the loading of the head experienced by athletes during impacts. However, only in the last decade has instrumented equipment been used to collect large amounts of data for full sporting teams over entire seasons. Instrumented helmets and skullcaps have been used in American football and ice hockey, whilst instrumented headgear and headbands have been used in boxing and soccer. Instrumented mouthguards and skin patches have been developed for use in contact and collision sports that do not require wearing helmets or headgear such as soccer, rugby league, rugby union, and Australian football. The main advantage of using instrumented equipment is the ability to estimate the head impact kinematics of human subjects *in vivo*.

The objective of the current study is to discuss the development, validity, and potential of different instrumented equipment: helmets, headgear, headbands, skullcaps, skin patches, and mouthguards.

2. Development of Instrumented Equipment

In 1961, the Committee on the Medical Aspects of Sports of the American Medical Association was concerned with the incidence of head injuries in American football and suggested the gathering of head impact data [1]. In response, Aagaard and Du Bois [2] instrumented a suspension helmet with a triaxial accelerometer, which was able to telemeter impact data for a linebacker during a professional American football game. Reid et al. [3] further developed the telemetry system and collected head impact data for an American football player during collegiate games over several seasons. Similarly, Moon et al. [4] developed an instrumented headband, which was worn underneath the helmet. However, both studies recorded peak linear accelerations in excess of $1000\,g$ [3, 4], which were much greater than contemporaneous head injury tolerance limits [5, 6]. Reid

et al. [7–10] revised the instrumented helmet system by mounting the accelerometers on the suspension system in an attempt to obtain more representative data; however, no-injury impacts of up to 400 g were still being recorded, which is higher than the pass criteria of a modern American football helmet standard [11]. One instrumented helmet captured a concussion, which had a peak linear acceleration of 188 g, and another instrumented player reported feeling "fuzzy," which may have been considered a concussion under the current sports-related consensus definition [12], from an impact with unremarkable peak linear acceleration. Early attempts to collect data during American football games using instrumented helmets were considered largely unsuccessful due to the technical difficulties associated with safely obtaining accurate measurements [13]. Concomitantly, instrumented mouthpieces were used to measure kinematic data in early human volunteer sled test studies investigating injury tolerance limits for automotive and aerospace applications [14–27].

A decade later, Morrison [28] attempted to use instrumented helmets to investigate head impacts in American football; however, similarly to previous studies [3, 4, 7–10], unreasonably high peak linear accelerations in excess of 500 g were recorded. Instrumented helmets were not used to gather head impact data again until Naunheim et al. [29] recorded the peak linear head accelerations of American football and ice hockey players during high school games. Peak linear head accelerations of up to 120 g and 150 g were measured for the American football and ice hockey players, respectively; however, no concussions were reported.

3. Instrumented Helmets and Headgear

In 2003, Greenwald et al. [30] developed the Head Impact Telemetry (HIT) System, which uses a novel computational algorithm to process data from a nine-accelerometer array incorporated into a helmet, allowing continuous sideline monitoring of head impacts in real-time [31, 32]. Pendulum impact testing using a Hybrid III anthropomorphic test device (ATD) head-neck system was performed to evaluate the accuracy of the HIT System [33], from which the mean error for linear acceleration was 4%; however, angular accelerations had a mean error of 17%. Manoogian et al. [34, 35] investigated 50 helmet-to-helmet impacts using two HIT System helmets mounted on Hybrid III ATD head-neck systems in a pendulum arrangement and found peak linear acceleration of the head to be less than 10% of peak linear acceleration of the helmet, which may explain the high results of early helmet instrumentation studies [3, 4, 7–10, 28]. Manoogian et al. [34, 35] reported a good agreement between the measured accelerations of the HIT System and the Hybrid III ATD headform. Similar validation assessment studies were also performed on HIT Systems, which had been modified for use in boxing [36, 37] and soccer [38].

To address the limitation of angular acceleration estimation accuracy, Chu et al. [109] developed a revised computational algorithm, which iteratively optimised the equations of motion for a head impact to determine the full head kinematics with six degrees-of-freedom (6DOF). The revised algorithm was implemented in a revised system, which comprised 12 uniaxial accelerometers arranged in orthogonal pairs, tangential to the skull, in six locations within the helmet. In a validation assessment of the 6DOF system, Rowson et al. [94] used a linear impactor to impact an American football helmet mounted on a Hybrid III ATD head-neck system. Mean errors for peak linear and angular acceleration were reportedly 1% and 3%, respectively. Although the revised system offered more accurate data for angular acceleration in comparison to the HIT System, prohibitive costs limited the widespread implementation of the 6DOF system [110].

Beckwith et al. [95] also used a linear impactor to deliver impacts to an American football helmet, which was instrumented with the HIT System, mounted on a Hybrid III ATD head-neck system. The HIT System was found to overestimate linear acceleration and underestimate angular acceleration of the Hybrid III ATD headform by 1% and 6%, respectively.

Allison et al. [96] evaluated the accuracy of the HIT System for ice hockey helmets. A linear impactor was used to impact an instrumented helmet, which was mounted on a Hybrid III ATD head-neck system, at speeds ranging from 1.5 to 5.0 m/s to several sites: front, rear, side, oblique-rear, and oblique-front. Initially, the effect of the interface between the helmet and the ATD headform was investigated using three interface conditions: nylon skull cap to mimic previous validation assessment studies [94, 95], dry human hair wig, and wet human hair wig. The latter was chosen by Allison et al. [96] as the most realistic interface condition. The HIT System algorithm identified almost a fifth (19%) of all impacts as perturbations, especially frontal impacts to the facemask, and removed such data. For the remaining impacts, peak linear and angular accelerations were found to strongly correlate with the reference data recorded by the ATD headform; however, correlations varied with impact location and the error associated with the HIT System data was found to be greater than previously reported for American football helmets [94]. Wilcox et al. [111], which included developers of the HIT System, criticised the protocol used by Allison et al. [96] for not being representative of on-ice conditions; however, Arbogast et al. [112] defended the protocol used by Allison et al. [96], which was chosen to mimic previous HIT System studies for boxing headgear [37] and American football helmets [94, 95].

Similarly for American football helmets, Jadischke et al. [100] used a linear impactor to investigate the accuracy of the HIT System for two helmet sizes: medium and large. For the large size helmet tests, the front, rear, and sides of the helmet shell were impacted at speeds of approximately 9.3 m/s. For the medium size helmet tests, various helmet shell sites were impacted at speeds ranging from 5.0 to 11.2 m/s. Root-mean-square (RMS) errors of HIT System linear and angular accelerations from the ATD headform data for the large size helmet were 18% and 66%, respectively, and for the medium size helmet were 18% and 20%, respectively. The medium size helmet was also impacted to face mask sites,

for which RMS deviations of HIT System linear and angular accelerations from the ATD headform data were 148% and 71%, respectively. Jadischke et al. [100] also investigated the pressure exerted by an American football helmet on the head of a volunteer high school player using a nylon skull cap. A medium size helmet mounted on a 50th Hybrid III ATD headform was found to exert peak pressures of 93 kPa, which were in excess of the discomfort pressure of 69 kPa reported by volunteer high school players. Previous validation assessment studies had used medium size helmets [94, 95], which Jadischke et al. [100] considered too tight in comparison to comfortable helmet pressures. Similar to the study of Allison et al. [96], the protocol of Jadischke et al. [100] was also criticised for not being representative of on-ice conditions.

Siegmund et al. [108] assessed the validity of the HIT System using a linear impactor to impact a mandibular load-sensing headform (MLSH) [113], which was wearing an American football helmet and mounted on a Hybrid III ATD neck. For peak linear acceleration and peak angular acceleration, the HIT System did not achieve level 1 validity, which was arbitrarily defined as "an average intercept and slope that were not statistically different from zero and one, respectively, for all impact sites combined".

For over a decade, numerous studies have used the HIT System to collect a large amount of head kinematic data from American football and ice hockey players (Table 1) [114], which have been used to further the understanding of concussion and investigate injury tolerance criteria [115]. The HIT System has also been used to monitor the head impacts of boxers [36, 37, 43] and female ice hockey players [52, 66, 83, 116].

The GForceTracker (GFT) comprises a triaxial accelerometer and gyroscope and, similar to the HIT System, allows continuous collection of head impact data in real-time [117]. In contrast to the HIT System, a GFT unit is attached to the helmet, which allows integration with a helmet of choice across a range of sports. During an impact, the GFT samples linear acceleration and angular velocity at 3000 Hz and 800 Hz, respectively, with the angular velocity signal passing through a low-pass filter with a cut-off frequency of 100 Hz.

Allison et al. [101] evaluated the accuracy of the GFT using a linear impactor, at speeds ranging from 1.5 to 5.0 m/s, to impact a hockey helmet mounted on a Hybrid III ATD head-neck system at various sites: facemask, side, rear-oblique, and rear. Relative to the peak linear acceleration data from the Hybrid III ATD headform, the raw data from the GFT demonstrated large differences of up to 150%, which was attributed to the lack of algorithm to transform the data to the centre of gravity of the head. When logistic regression was used to account for impact direction, mean absolute errors of up to 15% were obtained, which varied by helmet brand, impact direction, and sensor location, but not impact severity. In contrast, relatively small raw data differences of up to 15% were reported for angular velocity and mean absolute errors of less than 10% were obtained after logistic regression was used to account for impact direction. Mean absolute errors for angular velocity did not vary substantially

by helmet brand, impact direction, sensor location, or impact severity. Allison et al. [101] recommended that helmet brand-specific correction algorithms be developed to transform the raw linear acceleration data obtained from the GFT to represent the kinematics of the centre of gravity of the head.

In a similar study, Campbell et al. [104] used a linear impactor to impact an American football helmet mounted on a Hybrid III ATD head-neck system to assess the accuracy of the GFT. Impact speeds ranged from 3.0 to 5.5 m/s and various helmet locations were impacted: facemask, front, front-oblique, side, and rear. A correction algorithm was developed and used to predict the kinematics at the centre of mass of the head. Campbell et al. [104] found a strong correlation ($R^2 = 0.97$) between the peak linear accelerations measured by the GFT with the correction algorithm applied and the Hybrid III ATD headform data. A strong correlation ($R^2 = 0.94$) was also found between raw peak rotational velocity measured by the GFT and the Hybrid III ATD headform data. Campbell et al. [104] supported the conclusions of Allison et al. [101] regarding helmet brand-specific correction algorithms.

Certification of instrumented helmets is a contentious issue [118]. In 2013, National Operating Committee on Standards for Athletic Equipment (NOCSAE) published a press release stating that American football helmets with additional third-party products, such as impact sensors, which were not affixed during standards testing, voided the certification of compliance with the standard [119]. Several months later, NOCSAE published a clarification stating that helmet manufacturers were required to decide whether additional third-party products voided the certification of their helmets [120].

4. Instrumented Mouthguards

Half a century ago, instrumented mouthpieces were used to measure kinematic data in early human volunteer sled test studies [14–27]. More recent studies have used instrumented mouthpieces, which resemble mouthguards used for orofacial protection in sports, to investigate head kinematics during soccer heading with [121–123] and without [124–126] head protection; however, such devices were hardwired and not suitable for in-game situations. Higgins et al. [127] conducted impact drop tests to compare acceleration data from an instrumented mouthpiece and helmet with data from a modified NOCSAE headform. A significant relationship was observed between mouthpiece and headform acceleration; however, helmet acceleration was not significantly associated with headform acceleration.

More recently, Paris et al. [128] instrumented a custom acrylic mouthguard with a single dual-axis accelerometer, which was able to wirelessly transmit linear acceleration data. Kara et al. [129] further developed the instrumented mouthguard design of Paris et al. [128] to incorporate an array of three accelerometers so that angular acceleration, in addition to linear acceleration, was able to be measured. Six heading events were conducted, which all involved the same female subject. The device was suggested as a

TABLE 1: HIT System studies of male athletes during training and games.

Study	Season(s)	Sport	Level (age)	Players	Impacts[a]	Concussions
Duma et al. [39, 40]	2003	American football	Collegiate	38	3312	1
Brolinson et al. [41]	2003-2004	American football	Collegiate	52	11,604	3
Gwin et al. [42]	2005	Ice hockey	Collegiate	4	1875	0
Greenwald et al. [43]	2003–2006	American football	Collegiate	423	249,613	11
	2005	Ice hockey	Collegiate	5	5463	0
Funk et al. [44]	2003–2005	American football	Collegiate	56	22,704	3
Funk et al. [45]	2003–2007	American football	Collegiate	64	27,319	4
Guskiewicz et al. [46]	2004–2006	American football	Collegiate	88	104,714	13
Mihalik et al. [47]	2005-2006	American football	Collegiate	72	57,024	0
Schnebel et al. [48]	2005	American football	High school	16	8326	3
			Collegiate	40	54,154	3
Greenwald et al. [49]	2004–2006	American football	High school	190	99,862	6
			Collegiate	259	190,054	11
Mihalik et al. [50]	2006/2007	Ice hockey	Youth (13 years)	14	4543	0
Beckwith et al. [51]	2005–2008	American football	High school, collegiate	52	71,390	55
Brainard et al. [52]	2008-2009	Ice hockey	Collegiate	10	Not reported	0
Broglio et al. [53]	2007	American football	High school	35	19,224	0
Cubos et al. [54]	2006/2007	Ice hockey	Youth (13-14 years)	13	1137	0
Duma and Rowson [55]	2003–2008	American football	Collegiate	64	71,300	6
Gwin et al. [56]	2005-2006	Ice hockey	Collegiate	12	4393	0
Rowson et al. [57]	2007	American football	Collegiate	10	1712	0
Broglio et al. [58]	2005–2008	American football	High school	78	54,247	13
Crisco et al. [59]	2007	American football	Collegiate	188	Not reported	0
Broglio et al. [58]	2007–2010	American football	High school	95	101,994	20
Mihalik et al. [60]	Not reported	Ice hockey	Youth (14 years)	16	Not reported	0
Reed et al. [61]	2006-2007	Ice hockey	Youth (13-14 years)	13	1821	0
Kimpara et al. [62]	2007-2008	American football	Collegiate	19	4709	0
Beckwith et al. [63, 64] Beckwith et al. [65]	2005–2010	American football	High school, collegiate	1208	Not reported	105
Brainard et al. [66]	2008/2009-2009/2010	Ice hockey	Collegiate	44	15,281	0
Breedlove et al. [67]	2009-2010	American football	High school	24	Not reported	4
Daniel II et al. [68, 69]	Not reported	American football	Youth (7–8 years)	7	748	0
Duhaime et al. [70]	Not reported	American football, ice hockey	Collegiate	450	486,594	48
Funk et al. [71]	2006–2010	American football	Collegiate	98	37,128	8
Gysland et al. [72]	Not reported	American football	Collegiate	46	Not reported	0

Table 1: Continued.

Study	Season(s)	Sport	Level (age)	Players	Impacts[a]	Concussions
Mihalik et al. [73]	Not reported	Ice hockey	Youth (13–16 years)	52	12,253	0
Ocwieja et al. [74]	2010	American football	Collegiate	46	7992	0
Rowson et al. [75]	2007–2009	American football	Collegiate	335	300,977	57
Rowson et al. [76] Urban et al. [77]	2012	American football	Youth (7–12 years), middle school, high school	112	34,603	8
Cobb et al. [78]	Not reported	American football	Youth (9–12 years)	50	11,978	4
Talavage et al. [79]	2009	American football	High school	21	15,264	4
Urban et al. [77, 80] Stitzel et al. [81]	Not reported	American football	High school	40	16,502	5
Daniel II et al. [82]	2012	American football	Youth (12–14 years)	17	4678	1
Stitzel et al. [81]	Not reported	American football	Youth (9–12 years)	36	9772	0
Wilcox et al. [83]	2009/2010–2011/2012	Ice hockey	Collegiate	41	19,880	0
Young et al. [84]	2011-2012	American football	Youth (7-8 years)	19	3059	0
Cummiskey [85]	2013	American football	High school	12	224	0
Munce et al. [86]	Not reported	American football	Youth (11–13 years)	22	6183	1

[a]Number of recorded impacts that surpass a predefined minimum linear acceleration threshold.

potential tool for the assessment of concussion during game play.

The X2 Impact mouthguard is a device instrumented with a triaxial accelerometer and gyroscope, which samples linear acceleration and angular velocity at 1 kHz and 800 Hz, respectively, during an impact. Angular velocity is interpolated to 1 kHz, filtered, and differentiated to generate angular acceleration. Using a similar design to the X2 Impact mouthguard, Camarillo et al. [97] evaluated the accuracy of an instrumented mouthguard for measuring kinematics of the head during impact. A custom ATD headform mounted on a Hybrid III ATD neck and wearing an American football helmet was impacted at various sites using a linear impactor. The normalised RMS errors for impact kinematic profiles were approximately 10% for peak linear acceleration, angular acceleration, and angular velocity. An impedance-based saliva sensor is incorporated for on-field use to determine actual impact events when the mouthguard is present in the mouth. King et al. [130] monitored head impacts of 38 New Zealand amateur rugby union players using the X2 Impact mouthguard during the 2013 season; however, no concussions were recorded. Siegmund et al. [108] assessed the validity of the X2 mouthguard using a linear impactor to impact a mandibular load-sensing headform (MLSH) [113], which was wearing an American football helmet and mounted on a Hybrid III ATD neck. Similar to the HIT System results from the same study, the X2 Impact mouthguard did not achieve level 1 validity for peak linear acceleration and peak angular acceleration.

Kuo et al. [107] investigated the effect of mandible constraints on the accuracy of an instrumented mouthguard for helmeted ATD and cadaver tests. RMS errors of 40% and 80% for angular velocity and acceleration, respectively, were found for the worst-case scenario of the unconstrained cadaver mandible; however, such errors could be mitigated to below 15% by isolating sensors from mandible loads. Hernandez et al. [131] used instrumented mouthguards to monitor head impacts of American football players during collegiate games and training, the data from which informed ATD reconstructions. In addition, Hernandez et al. [132] also monitored head impacts to boxers and mixed martial artists, which, in combination with video analysis, enabled such impacts to be reconstructed using a finite element human head model [133].

Wu et al. [134] developed an instrumented mouthguard with a triaxial high-range accelerometer and gyroscope. The device also incorporated infrared proximity sensing to determine if the mouthguard is worn on the teeth. Frequency domain features of linear acceleration and rotational velocity measured by a Hybrid III headform during impacts at speeds ranging from 2.1 to 8.5 m/s delivered by a linear impactor were used to train a support vector machine classifier. In a subsequent study, Wu et al. [106] assessed the validity of the instrumented mouthguard for soccer heading impacts by tracking fiducial grids with dual high-speed video. The instrumented mouthguard was worn by the volunteer during the heading of a soccer ball, which was projected at a speed of 7 m/s. Compared to the video-tracked kinematics in the sagittal plane, the instrumented mouthguard had RMS

errors of 16%, 18%, and 12% for peak anterior-posterior linear acceleration, peak inferior-superior linear acceleration, and peak angular velocity in the sagittal plane, respectively.

Contemporaneously, the Cleveland Clinic developed the Intelligent Mouthguard (IMG) [98, 99, 102], comprising a triaxial accelerometer and gyroscope, which is capable of sampling up to 4 kHz. A drop tower was used to validate the sensors used in the IMG for linear and angular accelerations ranging within 10–174 g and 0.85–10.00 krad/s^2, respectively, with impact durations of 4.6–31.8 ms. The IMG underestimated the reference linear and angular accelerations by 3% and 17%, respectively. In addition, validation of the IMG was performed by impacting a modified Hybrid III ATD head, which was wearing either an American football helmet or boxing headgear, with a linear impactor at speeds of up to 8.5 m/s [102]. The accelerations recorded by the IMG correlated well with the headform data (R^2 = 0.99) for both the American football helmet and boxing headgear tests. Bartsch et al. [102] instrumented two collegiate American football players and four amateur boxers during competition; however, no concussions were recorded.

5. Instrumented Skin Patches

The X2 X-Patch is a small microelectromechanical system, worn over the left or right mastoid process, which comprises a triaxial accelerometer and gyroscope [135]. The raw accelerometer data is transformed to the centre of gravity of the head using a rigid body transformation for linear acceleration and a five-point stencil for rotational acceleration. During an impact, the X-Patch samples linear acceleration and angular velocity at 1 kHz and 800 Hz, respectively. Several studies have used the X-Patch to record kinematic data during training sessions and competition games for male American football, female soccer and youth rugby (Table 2).

Nevins et al. [105] assessed the validity of the X-Patch using a Hybrid III ATD head-neck system mounted on a low friction sled. The headform was impacted to the chin and forehead in different orientations by pneumatically projected softballs, lacrosse balls, and soccer balls at speeds ranging from 10 to 31 m/s. Peak linear acceleration measured by the X-Patch displayed reasonable agreement with the Hybrid III ATD headform for the lacrosse and soccer balls. However, peak linear acceleration was underestimated by the X-Patch for softball impacts as was angular acceleration for all three sports balls. Nevins et al. [105] suggested that the poor agreement between the X-Patch and Hybrid III ATD headform for certain conditions was attributable to the relatively low sampling frequency of the former.

Kerr et al. [136] evaluated the effectiveness of the Heads Up Football (HUF) programme using the X-Patch to monitor head impacts of youth football players: HUF participants, 38 players, 7 teams, 2 leagues; controls, 32 players, 8 teams, 3 leagues. Players participating in the HUF programme accumulated fewer impacts per athletic exposure than controls. Similarly, Swartz et al. [88] evaluated the effectiveness of

TABLE 2: X2 X-Patch studies of athletes during training and games.

Study	Season(s)	Sport	Level (age)	Gender	Players	Impacts[a]	Concussions
Kerr et al. [87]	2014	American football	Youth (8–15 years)	M	70	7478	6
Swartz et al. [88]	2014-2015	American football[b]	Collegiate	M	50	Not reported	Not reported
McCuen et al. [89]	Not reported	Soccer	High school, collegiate	F	43	Not reported	0
Morrison et al. [90]	2015	American football	Collegiate	M	10	Not reported	0
Stucker [91]	Not reported	Soccer	Collegiate	F	25	13,479	0
Cummiskey [85]	2013	American football	High school	M	15	231	0
King et al. [92]	2015	Rugby	Youth (8-9 years)	M	14	721	3
Svaldi et al. [93]	Not reported	Soccer	High school	F	14	Not reported	0

[a]Number of recorded impacts that surpass a predefined minimum linear acceleration threshold. [b]Helmetless training drills.

the Helmetless Tackling Training (HuTT) programme, which incorporates tackling drills without helmets and shoulder pads into training sessions, over the 2014 and 2015 collegiate seasons. Intervention and control groups, each comprising 25 American football players, were instrumented with the XPatch with the former participating in the HuTT programme. Swartz et al. [88] found that the intervention group experienced 28% less head impacts per athletic exposure compared to the control group. In another study, which varied the level of equipment worn by American football players, Reynolds et al. [137] instrumented 20 collegiate players with the X-Patch during training and games comprising the 2013 season. The type of equipment worn during each training session was found to be associated with different head impact profiles as mean peak linear and angular accelerations for helmet-only training sessions were significantly less than mean peak accelerations for half-pad and full-pad training sessions and competitive games.

In a soccer heading study, Wu et al. [106] assessed the validity of the X-Patch by tracking fiducial grids with dual high-speed video. The X-Patch was attached to the mastoid process of a volunteer during the heading of a soccer ball, which was projected at a speed of 7 m/s. Compared to the video-tracked kinematics in the sagittal plane, the X-Patch had RMS errors of 14% for peak anterior-posterior linear acceleration and 29% for both peak inferior-superior linear acceleration and peak angular velocity in the sagittal plane.

More recently, King et al. [92] used the X-Patch to monitor the magnitude, frequency, and location of head impacts to junior rugby union players in New Zealand over four consecutive matches. Of the 14 instrumented players, three were medically diagnosed as having sustained a concussion. The standardisation of reporting of head impact biomechanical data was suggested to enable accurate comparison across published studies.

6. Instrumented Skullcaps and Headbands

The Checklight is a sensor device, which is integrated into the rear of a skullcap [138]. Impact data is not provided; however, green, yellow, and red lights are triggered for "mild," "intermediate," and "severe" impacts, respectively.

Cummiskey [85] used an impulse hammer to impact an American football helmet, which was worn by a Hybrid III ATD over the Checklight skullcap. The red light was triggered by four impacts, the most severe of which corresponded to peak linear and angular headform accelerations of 123 g and 7660 rad/s^2, respectively. Bartsch et al. [102] used a Checklight skullcap in a validation assessment study; however, no results were reported. Harper et al. [139] monitored head impacts of youth and high school football players during training and games using the Checklight. Harper et al. [139] concluded that the Checklight has limited usefulness as it does not allow for real-time sideline data monitoring and threshold limits are unknown. In a soccer heading study, Wu et al. [106] assessed the validity of the Checklight skullcap sensor location by tracking fiducial grids with dual high-speed video. A 6DOF sensor device [134] was used in lieu of the Checklight sensor, which does not allow raw data extraction. Compared to the video-tracked kinematics in the sagittal plane, the skullcap had RMS errors of 16% for peak anterior-posterior linear acceleration and 13% for both peak inferior-superior linear acceleration and peak angular velocity in the sagittal plane.

Instead of traditional accelerometers, the Shockbox uses four binary force switches to measure differential voltage [140]. Foreman and Crossman [141] assessed the validation of the Shockbox by drop testing a rigid headform wearing an ice hockey helmet, which was instrumented with the device. Impact speeds ranged from 2.0 to 3.0 m/s at various helmet locations: front, front-oblique, side, rear-oblique, and rear. An aggregate difference of 9% between the Shockbox and headform data was reported. Cuminsky [85] used an impulse hammer to impact an American football helmet, which was worn by a Hybrid III ATD over the Shockbox headband. Peak linear acceleration from the Shockbox was compared to the headform data and RMS errors of 92–298% were found for seven impact locations. Wong et al. [142] instrumented the helmets of 22 youth American football players with the Shockbox device to monitor head impacts during the 2012 season. Other unpublished studies have used Shockbox to monitor head impacts in American football [143] and ice hockey [144, 145].

The SIM-G is a head impact sensor device, which comprises a triaxial gyroscope and two triaxial accelerometers,

TABLE 3: Validation studies of head instrumentation devices.

Study	Device	Sport	Method	Headform	Impact speed [m/s]	Error data — Linear	Error data — Angular
Crisco et al. [33]	HITS	American football	Pendulum	HIII head-neck	3.0–7.0	PLA: average error of 4% PAA: average error of 17%	
Manoogian et al. [35]	HITS	American football	Pendulum	HIII head-neck	2.0–5.0	HITS and HIII head measure similar linear acceleration responses	
Beckwith et al. [37]	HITS	Boxing	Pendulum	HIII head-neck	3.0–7.0	PLA: underestimate by 2%; $R^2 = 0.91$	PAA: overestimated by 8%; $R^2 = 0.91$
Hanlon and Bir [38]	HITS	Soccer	Ball canon	HIII head-neck	8.0–12.0	PLA: $R^2 = 0.34$	PAA: $R^2 = 0.57$
			Linear impactor	HIII head-neck	2.5–4.75	PLA: average error of -2 g; $R^2 = 0.94$	PAA: average error of -1721 rad/s^2; $R^2 = 0.92$
Rowson et al. [94]	6DOF System	American football	Linear impactor	HIII head-neck	Impact energy: 68–608 J	PLA: RMSE $= 12.5 \pm 8.3$ g; $R^2 = 0.88$	PAA: RMSE $= 907 \pm 685$ rad/s^2; $R^2 = 0.85$
Beckwith et al. [95]	HITS	American football	Linear impactor	HIII head-neck	4.4–11.2	PLA: $R^2 = 0.82$–0.99; overestimated by 1%	PAA: $R^2 = 0.42$–0.98; underestimated by 6%
Allison et al. [96]	HITS	Ice hockey	Linear impactor	HIII head-neck	1.5–5.0	PLA: average error of 18–31% (raw), 7–27% (adjusted); $R^2 = 0.81$–0.97	PAA: average error of 35–64% (raw), 13–38% (adjusted); $R^2 = 0.71$–0.94
Camarillo et al. [97]	MG	American football	Linear impactor	Custom head, HIII neck	3.0–8.5	PLA: RMSE $= 3.9 \pm 2.1$ g; NRMS $= 9.9 \pm 4.4\%$; $R^2 = 0.94$–0.98	PAA: RMSE $= 202 \pm 120$ rad/s^2; NRMS $= 9.7 \pm 7.0\%$, $R^2 = 0.61$–0.98; PAV: RMSE $= 1.0 \pm 0.8$ rad/s; NRMS $= 10.4 \pm 9.9\%$; $R^2 = 0.96$–0.99
Bartsch et al. [98], Aksu [99]	IMG	N/A	Drop	N/A	0.7–3.9	PLA: underestimated by 3%; $R^2 > 0.99$	PLA: underestimated by 17%; $R^2 = 0.98$
Jadischke et al. [100]	HITS	American football	Linear impactor	HIII head-neck	9.3; 5.0–11.2	PLA: RMSE $= 12$–23% PLA: RMSE $= 16$–190% PLA: RMSE $= 11$–22%	PAA: RMSE $= 30$–111% PAA: RMSE $= 51$–96% PAA: RMSE $= 10$–26%
Allison et al. [101]	GFT	Ice hockey	Linear impactor	HIII head-neck	1.5–5.0	PLA: $R^2 = 0.77$–0.99	PAV: $R^2 = 0.78$–0.99
Bartsch et al. [102]	IMG	American football	Linear impactor	HII head (modified), HII neck	2.0–8.5	PLA: $R^2 = 0.99$	PAA: $R^2 = 0.98$ PAV: $R^2 = 0.99$
		Boxing	Linear impactor	HII head (modified), HII neck	1.8–7.5	PLA: $R^2 = 0.99$	PAA: $R^2 = 0.98$ PAV: $R^2 = 0.98$
Triax Technologies [103]	SIM-G	N/A	Pendulum	NOCSAE head, Hybrid III neck	Not reported	PLA: $R^2 = 0.84$	PAA: $R^2 = 0.78$ PAV: $R^2 = 0.98$
Campbell et al. [104]	GFT	American football	Linear impactor	HIII head-neck	3.0–5.5	PLA: MAPE $= 26$–72% (raw), 2–8% (adjusted); $R^2 = 0.82$ (raw), 0.97 (adjusted)	PAV: MAPE $= 2$–16% (raw), 1–13% (adjusted); $R^2 = 0.94$ (raw), 0.96 (adjusted)

TABLE 3: Continued.

Study	Device	Sport	Method	Headform	Impact speed [m/s]	Error data — Linear	Error data — Angular
Cummiskey [85]	HITS	American football	Impulse hammer	HIII head-neck	Not reported	PLA: RMSE = 33–198%	PAA: RMSE = 27–209%
	X2 X-Patch	American football	Impulse hammer	HIII head-neck	Not reported	PLA: RMSE = 11–59%	PAA: RMSE = 11–350%
	Shockbox	American football	Impulse hammer	HIII head-neck	Not reported	PLA: RMSE = 92–298%	
	SIM-G	American football	Impulse hammer	HIII head-neck	Not reported	PLA: RMSE = 13–75%	
Nevins et al. [105]	X2 X-Patch	Soccer	Ball canon	HIII head-neck	13–27		
		Softball	Ball canon	HIII head-neck	10–20	PLA: underestimated by <25%	
		Lacrosse	Ball canon	HIII head-neck	27–31		
Wu et al. [106]	MG	Soccer	Ball canon	Human volunteer	7	PLA: $16 \pm 6\%$ (A-P); $18 \pm 10\%$ (I-S)	PAV: $12 \pm 7\%$ (sagittal)
	X2 X-Patch	Soccer	Ball canon	Human volunteer	7	PLA: $14 \pm 2\%$ (A-P); $29 \pm 30\%$ (I-S)	PAV: $29 \pm 9\%$ (sagittal)
	Checklight	Soccer	Ball canon	Human volunteer	7	PLA: $16 \pm 9\%$ (A-P); $13 \pm 5\%$ (I-S)	PAV: $13 \pm 11\%$ (sagittal)
Kuo et al. [107]	MG	American football	Drop	ATD (no mandible)	1.4–4.4	PLA: NRMSE > 20% for 8/162 drops; $R^2 = 0.85$–0.99	PAA: $R^2 = 0.62$–0.99; PAV: $R^2 = 0.84$–0.99
				ATD (loose mandible)	1.4–4.4	PLA: $R^2 = 0.62$–0.99	PAA: $R^2 = 0.56$–0.99; PAV: $R^2 = 0.65$–0.99
				ATD (clenched mandible)	1.4–4.4	PLA: $R^2 = 0.94$–0.99	PAA: $R^2 = 0.92$–0.99; PAV: $R^2 = 0.98$–0.99
				Cadaver	1.4–4.4	PLA: NRMSE > 20% for 26/108 drops	PAA: NRMSE = 80%; PAV: NRMSE = 40%
Siegmund et al. [108]	HITS	American football	Linear impactor	MLSH, HIII neck	3.6–11.2	PLA: mean coefficients of variation of $15 \pm 9\%$; mean relative error of $20 \pm 50\%$; MAPE $= 32 \pm 43\%$; $R^2 = 0.63$	PAA: mean coefficients of variation of $21 \pm 21\%$; mean relative error of $9 \pm 49\%$; MAPE $= 35 \pm 36\%$; $R^2 = 0.59$
	X2 MG	American football	Linear impactor	MLSH, HIII neck	3.6–11.2	PLA: mean coefficients of variation of $8 \pm 4\%$; mean relative error of $-8 \pm 14\%$; MAPE $= 12 \pm 10\%$; $R^2 = 0.90$	PAA: mean coefficients of variation of $23 \pm 9\%$; mean relative error of $26 \pm 61\%$; MAPE $= 41 \pm 51\%$; $R^2 = 0.56$

HITS: Head Impact Telemetry System. 12DOF: twelve degree-of-freedom measurement device. MG: mouthguard. IMG: Intelligent Mouthguard. GFT: GForceTracker. MMA: mixed-martial arts. HIII: Hybrid III. ATD: anthropomorphic test device. MLSH: modified load sensing headform. PLA: peak linear acceleration. PAA: peak angular acceleration. PAV: peak angular velocity. MAPE: mean absolute percentage error. A-P: anterior-posterior. I-S: inferior-superior.

high-g, and low-g, mounted on a headband [103]. The developers of the SIM-G, Triax Technologies, assessed the validation of the device using a pendulum to impact the NOCSAE headform in 11 locations. Peak angular velocity measured by the SIM-G correlated strongly with the headform data ($R^2 = 0.98$); however, correlations were not as strong for peak linear acceleration ($R^2 = 0.84$) and peak angular acceleration ($R^2 = 0.78$). Cummiskey [85] used an impulse hammer to impact an American football helmet, which was worn by a Hybrid III ATD over the SIM-G headband. Peak linear acceleration from the SIM-G was compared to the headform data and RMS errors of 18–75% were found for seven impact locations.

7. Other Instrumented Equipment

Circa 2000, instrumented earplugs were developed for motorsport drivers after it was shown that instrumented helmets moved relative to the head during collisions [146, 147]. Such ear-mounted devices were also tested by military cadets during boxing matches [148]. In addition to the HIT System studies in boxing [36, 37, 149, 150], instrumented gloves have also been used to estimate punch force in the laboratory [151, 152] and during boxing matches [138, 152, 153]. Boxing shirts have also been developed, which are instrumented and detect hits during amateur boxing matches [154–159].

In recent years, global positioning system (GPS) units are commonly worn by elite rugby league [160, 161], rugby union [162–165], and Australian football [166, 167] players; however, the validity of such microsensors to detect collisions has been questioned [168]. Another device provides video footage from a first-person perspective using rugby headgear instrumented with a video camera [169, 170]; however, due to rules regarding rugby headgear design [171], the primary application of such a device is as a training tool to assess performance [170].

8. Conclusion

Recent advances in technology have enabled the development of instrumented equipment: helmets, headgear, headbands, skullcaps, skin patches, and mouthguards. The current study was conducted to review the development, validity and potential of such instrumented equipment, which estimates the head impact kinematics of human subjects *in vivo*.

The HIT System is widely used; however, it is expensive and limited in that it can only be incorporated into particular helmets and headgear. Other head impact sensors are less expensive, such as the Checklight and Shockbox, and are commercially available despite the lack of validation. In contrast, the GForceTracker is continually undergoing validation assessments and is only currently available for use in research. For some devices, laboratory validation studies have found large discrepancies between device measurements and headform data (Table 3), especially for certain impact directions. Such discrepancies may be a result of nonrigid skull coupling for helmets, headgear, headbands, skullcaps,

and skin patches. Relatively small errors have been reported for instrumented mouthguards; however, constraint limitations have been identified with clenched teeth providing the most accurate results.

Over the past decade, instrumented equipment has recorded millions of impacts in the laboratory, on the field, in the ring, and on the ice. Instrumented equipment is not without limitations; however, *in vivo* head impact data is crucial to investigate head injury mechanisms and further the understanding of concussion.

Competing Interests

The author declares that they have no competing interests.

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