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Head-Impact–Measurement Devices: A Systematic Review

Kathryn L. O'Connor, MS*; Steven Rowson, PhD†; Stefan M. Duma, PhD†; Steven P. Broglio, PhD, ATC*

*NeuroTrauma Research Laboratory, University of Michigan, Ann Arbor; †School of Biomedical Engineering and Sciences, Virginia Tech, Blacksburg

Context: With an estimated 3.8 million sport- and recreation-related concussions occurring annually, targeted prevention and diagnostic methods are needed. Biomechanical analysis of head impacts may provide quantitative information that can inform both prevention and diagnostic strategies.

Objective: To assess available head-impact devices and their clinical utility.

Data Sources: We performed a systematic search of the electronic database PubMed for peer-reviewed publications, using the following phrases: *accelerometer and concussion*, *head impact telemetry*, *head impacts and concussion and sensor*, *head impacts and sensor*, *impact sensor and concussion*, *linear acceleration and concussion*, *rotational acceleration and concussion*, and *xpatch concussion*. In addition to the literature review, a Google search for *head impact monitor* and *concussion monitor* yielded 15 more devices.

Study Selection: Included studies were performed in vivo, used commercially available devices, and focused on sport-related concussion.

Data Extraction: One author reviewed the title and abstract of each study for inclusion and exclusion criteria and then reviewed each full-text article to confirm inclusion criteria. Controversial articles were reviewed by all authors to reach consensus.

Data Synthesis: In total, 61 peer-reviewed articles involving 4 head-impact devices were included. Participants in boxing, football, ice hockey, soccer, or snow sports ranged in age from 6 to 24 years; 18% (n = 11) of the studies included female athletes. The Head Impact Telemetry System was the most widely used device (n = 53). Fourteen additional commercially available devices were presented.

Conclusions: Measurements collected by impact monitors provided real-time data to estimate player exposure but did not have the requisite sensitivity to concussion. Proper interpretation of previously reported head-impact kinematics across age, sport, and position may inform future research and enable staff clinicians working on the sidelines to monitor athletes. However, head-impact–monitoring systems have limited clinical utility due to error rates, designs, and low specificity in predicting concussive injury.

Key Words: concussion, accelerometer, biomechanics

Key Points

- Head-impact sensors have limited applications to concussion diagnosis but may provide sideline staff with estimates of athlete exposure and real-time data to monitor players.
- Given that concussion risk is influenced by many factors in addition to impact biomechanics, viewing an athlete's head-impact data may provide context for the clinician working on the sidelines, but impact sensors should not replace clinical judgment.

In 2012, among the 23.6 million US youth athletes who were involved in organized athletics, 19% participated in collision sports, and 57% participated in contact sports.[1] Athletes who engage in collision or contact sports are at greater risk for concussion,[2] and an estimated 1.6 million to 3.8 million sport- and recreation-related traumatic brain injuries (TBIs) occur annually in the United States.[3] Concussive injuries compose 8.9% of all high school and 5.8% of all collegiate athletic injuries.[4] The true concussion rate is likely to be underestimated due to underreporting.[5–7]

Concussion is a brain injury resulting from a direct or indirect blow to the head, typically resulting in transient neurologic impairment and neuropathologic symptoms.[8] After a direct or indirect force to the head, the brain experiences a metabolic crisis due to disrupted ion flow across the neuron membranes and decreased adenosine triphosphate availability to correct the perturbed ion flow.[9] This metabolic disruption lasts about a week, mirroring the timeline of clinical symptom recovery.[10,11] In animal models, researchers[12] have demonstrated that a second concussion during this period of brain vulnerability creates greater metabolic and cognitive impairment for a longer time. Similar findings of vulnerability and impaired recovery have also been shown in male athletes.[13]

Removing injured athletes from participation close to the time of injury reduces the risk of secondary injury when they are vulnerable to the cumulative effect of concussions.[12–14] However, given underreporting,[5,7] transient symptoms,[15] delayed onset of symptoms,[16] and the few concussions occurring with loss of consciousness, concussions are often difficult to detect and diagnose.[17–19] Therefore, objective and quantitative diagnostic tools that are more sensitive and specific to concussive injury are needed. Researchers have investigated more objective measures for concussion diagnosis, including balance testing,[20] neuropsychological testing,[21] and advanced imaging.[22] Whereas these measures assess the clinical symptoms of concussion, no measure can identify the concussed athlete while on the field or be used as a preventive tool. Head-impact biomechanics have been investigated to determine the kinematic signature of a concussion.[23] If the kinematics of concussive injury are elucidated, clinicians may be able to more rapidly identify the concussed athlete and advise protective-equipment improvements or rule changes to mitigate concussion risk.

Impacts to the head cause a combined linear and angular acceleration of the skull.[24] These accelerations result in transient pressure gradients and strain fields within the soft tissue of the brain.[25] If the pressure gradients or strains exceed the tolerable limits of the brain tissue, injury occurs. It is impossible to directly measure the tissue-level response of the brain to impact in vivo. Instead, skull acceleration is measured as a correlate to the pressure and strain responses of brain tissues. *Acceleration* represents the rate of change in velocity, and in this review, we report resultant linear accelerations (LAs) and resultant angular accelerations (AAs) of the head. *Resultant LA* is the vector sum magnitude of the 3-dimensional LAs of the skull resulting from an impact. It is measured in gravitational units (*g*), which is equal to the acceleration due to gravity (approximately 9.81 m/s^2). Similarly, *resultant AA* is the vector sum of the 3-dimensional AAs of the skull resulting from an impact and is measured in units of radians per second squared. Resultant LA and AA are closely correlated with each other.[26,27] The relative magnitude of resultant LA and AA depends on the distance between the force vector and the center of gravity of the head. Vectors farther away from the center of gravity create greater resultant AA relative to resultant LA. Alternatively, force vectors in line with the center of gravity create greater resultant LA.[28] Given that few impacts are solely aligned with the head's center of gravity, most impacts comprise both linear and rotational components. Therefore, in some instances, lowering resultant LA magnitude lowers resultant AA.

Determining the mechanics of brain injury is not a recent area of investigation. In 1943, Holbourn[29] studied the mechanics of head injuries, focusing on rotational forces. In subsequent primate work, Ommaya[30] suggested that, whereas resultant AA may produce diffuse and focal injury, resultant LA produces only focal injury. Since then, resultant LA and AA have been posited to influence concussive injury[23,31,32] through pressure gradients and shearing stress, respectively.[25] Whereas debate exists about whether resultant LA, resultant AA, or combined resultant LA and AA influence concussion risk,[32,33] they likely do not occur in isolation.[24]

In addition to simple magnitudes, impact-severity measures quantify injury tolerance, and the original work in car impacts yielded the Wayne State Tolerance Curve (WSTC).[34] The objective of the WSTC was to inform protective material development by understanding the risk of skull fracture in moderate and severe TBI.[35] The concept behind the WSTC is that humans can tolerate larger acceleration magnitudes for shorter periods and smaller accelerations for longer periods. Building on the WSTC, additional impact-severity measures, including the Gadd Severity Index[36] (GSI) and Head Injury Criteria[37] (HIC), were developed to study moderate to severe TBI. Whereas the GSI was a good tool for estimating short-duration impacts (ie, focal brain injuries), it was not as good at estimating longer-duration injuries that are more indicative of diffuse brain injury.[38] The HIC aimed to correct these shortcomings by using the portion of the acceleration-time curve with the greatest GSI score. However, GSI and HIC are still limited when we evaluate impacts with long durations.[39] Therefore, a shorter window to evaluate the greatest GSI score is required. A 15-millisecond HIC window (HIC15) was selected on the basis of auto-industry work, with cadaver simulations of injury indicating that durations less than 15 milliseconds mimic the WSTC.[39] Given that 95% of recorded head impacts last between 5.5 and 13.7 milliseconds,[40] the HIC15 is commonly used in the concussion literature. The HIC15 indicates that 15 milliseconds is the selected time range for integrating the linear time curve.

Although the GSI and HIC were developed to evaluate the likelihood of skull fracture with moderate to severe TBI, Greenwald et al[40] developed the Head Impact Telemetry System (HITS, Simbex, Lebanon, NH) severity profile (HITsp) to provide an estimate associated more with mild than with severe TBI. It combines resultant LA, rotational acceleration, impact location, and impact duration using a weighted principal component analysis. The HITsp measure has been shown to be more sensitive for concussion than resultant LA, resultant AA, or HIC15.[40] *Sensitivity* represents the proportion of concussions correctly identified, and *specificity* represents the proportion of nonconcussive impacts correctly identified as nonconcussive. Greater proportions for sensitivity and specificity indicate that the kinematic measure can discern between concussive and nonconcussive events. Greenwald et al[40] compared the sensitivity and specificity of head kinematics (resultant LA, resultant AA, HIC15, and HITsp) for estimating concussive injury. The proportion of nonconcussive events identified as concussive (ie, false-positive events) at the 90% sensitivity level ranged from 3.26% to 16.40% for the 4 kinematic metrics. Angular acceleration identified the greatest proportion of nonconcussive events as concussive (16.40%) at 90% sensitivity.[40] Further work combining resultant LA and AA demonstrated that the proportion of nonconcussive events identified as concussive was 4.0% (90% sensitivity).[33] Whereas this proportion is seemingly low, the number of these false-positive impacts surpassed the number of concussive impacts. Despite the motivations behind the creation of the GSI, HIC, and HITsp, these severity measures, along with resultant LA and AA, have not been strong estimators of concussive injuries due to their high false-positive rates.

Most biomechanics research on brain injury has been limited to laboratory experiments focusing on moderate to severe injury. Concussion has been particularly challenging to study because human-volunteer experiments must be noninjurious, and human surrogates (eg, cadaver, dummy) do not produce the physiological signs and symptoms required to identify concussion. Recent computer and technologic advances, however, have enabled in vivo concussion studies to be conducted using impact-monitoring devices. In addition, numerous devices have been marketed to athletes, parents, and clinicians with the suggestion that these devices offer clinical utility. Therefore, the purpose of our systematic review was to supply clinicians with a comprehensive review of currently available devices and discuss their clinical utility and limitations. First, we summarize data collected in vivo across age levels and sports. Second, results from laboratory studies provide context on the utility and limitations of each device.

METHODS

Data Sources

A PubMed search was completed in March 2016 using the following phrases: *accelerometer and concussion*, *head impact telemetry*, *head impacts and concussion and sensor*, *head impacts and sensor*, *impact sensor and concussion*, *linear acceleration and concussion*, *rotational acceleration and concussion*, and *xpatch concussion*. In addition to the literature review, a Google search for *head impact monitor* and *concussion monitor* yielded 15 more devices (Table 1). The reference list of each included article was reviewed for relevant additional articles.

Selection Criteria

Our review followed the Preferred Reporting Items for Systematic Reviews and Meta-Analyses (PRISMA) guidelines for systematic reviews to meet currently approved standards.[41] Inclusion and exclusion criteria were set before the search. We included articles if they were written in English and if the investigators studied an athlete sample and used concussion-specific accelerometer(s) to measure head mechanics. We excluded articles if they were laboratory studies, review papers, or commentaries or if investigators included a nonathlete sample, used an accelerometer that was not targeted for concussion measurement, or included a device that was not commercially available. Using these inclusion and exclusion criteria allowed us to generate the list of in vivo studies to evaluate and present compiled head-impact values.

Data Extraction

One author (K.L.O.) reviewed the title and abstract of each study for inclusion and exclusion criteria. She further reviewed each full-text article to confirm inclusion criteria. Controversial articles were reviewed by all authors (K.L.O., S.R., S.M.D., S.P.B.) to reach consensus. Data were extracted from the "Results" sections of included studies. When available, total number of impacts; impacts per season; impacts per session; and means, standard deviations, medians, interquartile ranges, and 95th percentiles were extracted for resultant LA, resultant AA, and HITsp.

RESULTS

Study Selection

A total of 285 records were initially identified via PubMed; of these, 186 were unique. Each unique study was evaluated for inclusion and exclusion criteria, and 46 were screened out by review of the title and abstract. To the remaining 140, we added 5 articles from citation lists. The 145 articles were reduced to a final list of 61 articles after excluding 84: laboratory studies (n = 59), studies not related to sport-related concussion (n = 3), review or response studies (n = 13), studies of devices that were not concussion specific (n = 8), and studies of devices that were not commercially available (n = 1; Figure 1). Basic demographics of study participants, along with impact counts and magnitudes, were extracted for the HITS or 6-degrees-of-freedom (6DOF) systems (n = 53) and X2 Biosystems (X2 Biosystems, Inc, Seattle, WA; n = 8). Device information, metrics, validity, and limitations were reviewed for all devices.

Devices

Of the 61 studies included in our review, 87% (n = 53) used the standard or 6DOF HIT; 11% (n = 7), X-Patch (X2 Biosystems, Inc); and 2% (n = 1), X-Guard (X2 Biosystems, Inc; Table 1). All devices are compared in Table 1, and normative data from published peer-reviewed manuscripts are described for each device in this section and are listed in Tables 2 and 3.

Helmeted Devices. The most commonly used device was the Riddell (BRG Sports, Rosemont, IL) HITS device. The HITS device used 6 single-axis accelerometers that fit inside a Riddell helmet. Specifically tailored versions of the HITS device have been placed in boxing,[89] ice hockey,[16,42,44,45,77,78,80] soccer,[87] and snow sports[88] headgear. For the device to record an impact, 1 of the 6 accelerometers must exceed 14.4*g*, but some users have adjusted this threshold to 9.6*g*. Resultant LA is recorded at 1 kHz, with 8 milliseconds of preimpact data and 32 milliseconds of postimpact data (determined when the threshold is reached) composing a 40-millisecond acceleration-time trace. When recorded, the data are filtered to eliminate any impact in which the peak resultant LA did not exceed 10*g*, meaning that all impacts in the final database were greater than 10*g*. The final output includes peak resultant LA, resultant AA, 40-millisecond resultant LA time trace, location of impact (location bins, azimuth, and elevation), HIC15, GSI, and HITsp. Resultant AA values are regression-based estimates derived from the resultant LA.

In addition to the HITS device, researchers have implemented a 6DOF device[31,85] (n = 2) or used the 6DOF with the standard HITS device[26,50,86] (n = 3). The 6DOF device is part of the HITS and fits within Riddell Revolution (BRG Sports) helmets. Compared with the HITS devices, the 6DOF uses 12 single-axis, 250*g* iMEMS accelerometers (model ADXL193; Analog Devices, Norwood, MA) that are paired and placed so their axes are tangential to the skull. An algorithm computes resultant LA.[51] Distributions of rotational acceleration have been shown to agree between the 6DOF and HITS devices,[26] and a strong correlation has been observed between the Hybrid III head model (Humanetics Innovative Solutions, Plymouth, MI) and 6DOF for resultant LA ($R^2 = 0.88$) and resultant AA ($R^2 = 0.85$).[51]

Nonhelmeted Devices. The X2 X-Patch and X-Guard use a triaxial accelerometer and gyroscope. The X2 devices are either worn behind the ear (X2 X-Patch) or embedded in

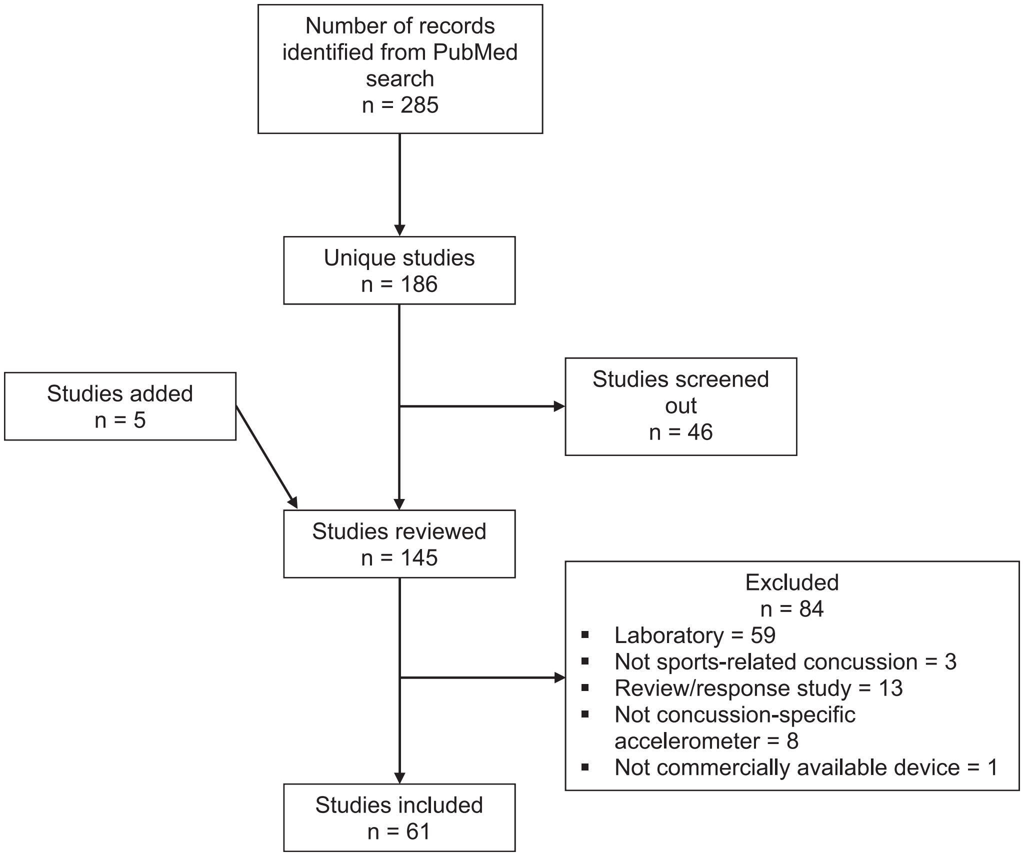

Figure 1. Systematic search.

a custom-formed mouthguard (X2 X-Guard). Resultant LAs are recorded at 1 kHz and resultant AAs are recorded at approximately 850 Hz, which is interpolated to match the resultant LA time sequence. The output includes peak resultant LA, peak resultant AA, HIC, location (azimuth and elevation), and direction of peak resultant LA. Compared with the HITS device, the X2 system supplies a 100-millisecond trace (10 milliseconds preimpact and 90 milliseconds postimpact) for each linear and angular sensor channel when an impact exceeds the 10*g* trigger.[90,98]

Device Limitations

Acceleration and loading of the head place strain on brain tissue that may cause concussive injury.[25] Given that directly measuring brain strain in vivo in humans is not possible,[99] impact sensors aim to couple with the skull and measure skull motion, an indirect measure of brain movement. Even in the ideal scenario with sensors affixed directly to the cranial bones, brain movement cannot be measured because the brain moves independently within the skull cavity.[23]

The helmeted design of HITS uses spring-loaded accelerometers to maintain contact with the skull as the helmet moves and deforms.[54] The HITS device assumes that the helmet and skull move as a single rigid body.[100] Consequently, HITS accuracy depends on good helmet fit because if the helmet is too loose, too much extraneous movement that is not coupled to the head will be present.[101] In this scenario, acceleration values may be overestimated because the helmet moves more than the skull.[102] The nonhelmeted designs of the X2 X-Patch and X-Guard attempt to address the limitations of helmeted designs and overcome the limitation of measuring head accelerations in nonhelmeted sports by placing sensors in the upper jaw[96] or behind the ear.[94] These designs have their own limitations related to skin motion, mouthguard fit,[96] saliva accumulation preventing data acquisition,[96] and a 50% error rate.[94]

Measurement Error With Helmeted Devices. Validation testing of the football HITS has shown that the resultant LA and rotational acceleration were within 4% of those values measured using a standardized headform,[54] but other researchers[47] noted the HITS may overestimate peak resultant LA by 8% and HIC by 23%. In more recent validity tests, Simbex, the manufacturer of the HITS, attempted to simulate National Football League impacts by using impact sites and velocities identified by the league. When compared with a Hybrid III headform, HITS overestimated resultant LA by 0.9% and underestimated peak resultant AA by 6.1%. Using specific impact locations and velocities possibly reduced the error compared with other studies. In another validation study, Rowson et al[26] investigated the HITS and 6DOF devices. Compared with the observations of Beckwith et al,[103] Rowson et al[26] found that resultant AA was overestimated, leading to a correction factor that has been applied to all HITS datasets since 2013. Error associated with individual data points is greater than aggregate distributions of the data. The pooled measurements are more representative of the distribution of resultant AA.[31] Siegmund et al[98] recently reported that

Table 1. Head-Impact Devices Extended on Next Page

Device by Company	Components	Real Time?	Output	Minimum Recording Threshold
Riddell[a]				
Riddell Sideline Response System	6 Single-axis accelerometers	Yes	Linear and angular acceleration, 40-ms linear-acceleration time curve, location (azimuth, elevation), direction, date, time, duration, Head Injury Criteria, Gadd Severity Index, and Head Impact Telemetry severity profile	User defined
Riddell Insight	5-Zone sensor pad	Yes	Head Impact Telemetry severity profile, identifies top 1%, 7-d accumulation of substantial impacts	Proprietary
X2 Biosystems[b]				
X-Patch	3-Axis accelerometer and gyroscope	Yes	Linear and angular acceleration, angular velocity, peak angular acceleration, and 100-ms linear- and angular-acceleration curves	User selected
X-Guard	3-Axis accelerometer and gyroscope	Yes	Linear and angular acceleration, Head Injury Criteria, location (azimuth, elevation), and 100-ms linear- and angular-acceleration curves	User selected
i1 Biometrics[c]				
Shockbox	4 Unidirectional, orthogonally placed force switches	Yes	Linear acceleration, force estimate, direction, date, time, and hit count	User selected
Vector	3-Axis accelerometer and gyroscope	Yes	Linear and angular acceleration	User selected
Brain Sentry[e]				
Brain Sentry	Not provided	Yes	Hit counts (wk and y) and alert light	20*g*
Jolt[f]				
Jolt	Not provided	Yes	Indicator light and vibration (3 levels)	Not provided
Head Case[g]				
Head Case	Not provided	No, up to 10 h after event	Force and Head Case *g* force	20*g*
Reebok[h]				
Checklight	Accelerometer and gyroscope	Yes	Total No. of impacts in each zone and indicator light (3 zones)	Not provided
Triax[i]				
SIM-P (individual) SIM-G (team)	3-Axis accelerometer and gyroscope	Yes	*g* Force, No. of impacts, and linear and angular acceleration	Range, 3*g*–150*g*
GForce Tracker[j]				
GForce Tracker	3-Axis accelerometer and gyroscope	Yes	Linear acceleration, angular velocity, 40-ms data capture, location (azimuth, elevation), Head Injury Criteria, and No. of impacts	User selected

Table 1. Extended From Previous Page, Continued on Next Page

Database	Alert	Distance	Location	Sports	Cost	Peer Reviewed?
Yes	Page alert if impact exceeds set threshold	50 yd (45 m)	In helmet	Football	$1000–$3000	Yes
Yes	Page alert if impact exceeds set threshold	50 yd (45 m)	In helmet	Football	$150/device, $200 alert monitor	No
Yes	Alert to injury-management software	Not provided	Behind ear	All	$150/device	Yes
Yes	No	NA	In mouthguard	All	Not available	Yes
Yes	Page telephone via Bluetooth[d]	Not provided	Helmet	All helmeted sports	$149.95/device	Yes
Yes	Page telephone or computer	Not provided	In mouthguard	Helmeted sports with face mask	NA	No
No	Light indicates the hardest 1.5% of hits	NA	Outside back of helmet	All helmeted sports	$75/device	No
Yes	Smartphone via Bluetooth[d]	200 yd (180 m)	Clips to headband or helmet at side of head	All	$99/device	No
Yes	E-mail when substantial increase in impact occurs (>50%)	Not provided	Helmet	All helmeted sports	$99.95/device	No
No	Light: green indicates low impact; yellow, moderate; and red, severe	NA	Skullcap	All	$99.97/cap	No
Yes	Bluetooth to iOS devices only, text or e-mail alert for impact above user set threshold[d]	150 yd (135 m)	Headband or skullcap	All	$189.99/individual SIM-P device	No
Yes	Alarm and flashing light-emitting diode when impact above set threshold	Not provided	Helmet	All helmeted sports	$150/device, $8/mo software	No

Table 1. Continued From Previous Page

Device by Company	Components	Real Time?	Output	Minimum Recording Threshold
Integrated Bionics, LLC[k]				
HeadsUp	Not provided	Yes	Hit count and training intensity	Not provided
Marucci Sports[l]				
BodiTrak Head Health Network	Accelerometer, gyroscope, and thermometer	Yes	No. of impacts, impact location, and helmet fit	Not provided
Linx[m]				
Impact Assessment System	3-Axis accelerometer and gyroscope	Yes	Impact Assessment System Score (proprietary) and indicator light (3 zones)	Not provided
Force Impact Technology[n]				
FITGuard	Not provided		Peak linear and angular acceleration and duration	User-selected or predetermined threshold based on age, weight, and sex
Archetype[o]				
PlayerMD	6-Degree-of-freedom sensor array	Yes	Linear and angular acceleration, No. of impacts, duration, and body temperature	15*g*
CSx[p]				
Hub	Not provided	Yes	Linear and angular acceleration and cumulative impact forces	Not provided
Samsung[q]				
brainBAND	Not provided	Yes	Linear acceleration and indicator light (3 zones)	Not provided

Abbreviation: NA, not applicable.
[a] BRG Sports, Rosemont, IL.
[b] X2 Biosystems, Inc, Seattle, WA.
[c] i1 Biometrics, Kirkland, WA.
[d] Bluetooth SIG, Inc, Kirkland, WA.
[e] Brain Sentry, Bethesda, MD.
[f] Jolt Athletics Inc, Boston, MA.
[g] Head Case LLC, Lake Forest, IL.
[h] Adidas AG, Herzogenaurach, Germany.
[i] Triax Technologies, Inc, Norwalk, CT.
[j] GForceTracker Inc, Richmond Hill, ON, Canada.
[k] Houston, TX.
[l] Marucci Sports, LLC, Baton Rouge, LA.
[m] Linx IAS Impact Assessment System, Blackbox Biometrics, Inc, Rochester, NY.
[n] Force Impact Technologies, Los Angeles, CA.
[o] Archetype, Inc, Pinson, AL.
[p] CSx Systems Ltd, Auckland, New Zealand.
[q] Samsung Electronics, Seoul, South Korea.

the HITS measures impacts to the front most accurately and performs worse when measuring impacts at the crown of the head. They observed that 55 of 64 impacts were in the direction opposite the actual impact direction.[98] However, the directional concerns may be attributed to the Riddell Revolution Speed helmet, given that the high rate of misclassification was eliminated when using the Riddell Revolution (the predecessor of the Revolution Speed).[98]

In a single study, Allison et al[104] attempted to validate the ice hockey HITS device. The error rates for the ice hockey HIT system ranged from 7% to 18% for resultant LA and 12% to 27% for resultant AA. These error rates were from data calibrated with regression equations, which estimated the data error at 18% to 31% for resultant LA and 35% to 64% for resultant AA.[104]

Measurement Error With Nonhelmeted Devices. The X-Patch also has measurement error as high as 50% for peak resultant LA and AA, suggesting that a 40*g* acceleration event will result in a reading between 20*g* and 60*g*.[94] Recently, Press and Rowson[93] examined the accuracy of the X-Patch compared with video-recorded impacts among collegiate soccer players. Whereas video captured 1703 confirmed head impacts, the X-Patch recorded 8999 impacts. The X-Patch recorded 7536 false-positive and 1463 true-positive impacts, yielding a positive predictive value of 16.3%. Error rates should be

Table 1. Extended From Previous Page

Database	Alert	Distance	Location	Sports	Cost	Peer Reviewed?
Yes	Light-emitting diode	NA	Headband or armband	All	$150/device	No
Yes	Alarm via Bluetooth when impact above set threshold[d]	Not provided	Helmet	Football	Contact directly	No
Yes	Alarm via Bluetooth and flashing light-emitting diode when impact above set threshold[d]	Not provided	Headband or skullcap	All	Coming soon	No
Not provided	Alarm via Bluetooth and flashing light-emitting diode when impact above set threshold[d]	Not provided	Mouthguard	All	Preorder, $99.99/ device	No
Yes	Alert when exceeds personalized threshold or in danger of overheating	0.5 mi	Headband or skullcap	All	$180/device	No
Yes	Visual reports	Not provided	Not provided	Rugby	Coming soon	No
Not provided	Alerts coaches, referee, and medical staff	Not provided	Headband	Rugby	Coming soon	No

considered when evaluating impact magnitudes[94] and frequencies.[93]

Recording Threshold

Recording thresholds for HITS literature varied from 9.6*g* to 15*g*. The default trigger is 14.4*g* on any single accelerometer, which does not reflect the resultant head acceleration. Although many researchers have suggested a 10*g* minimal value, it is most likely not the trigger value; rather, the HITS software filters all peak resultant LAs that are less than 10*g*. Consequently, the 10*g* reported values most likely refer to the filtering process and not the minimal trigger value of the single accelerometer. One needs to know the minimal triggering value to compare impact magnitudes across studies. King et al[105] evaluated how varying recording thresholds would change the head-impact data. By increasing the recording threshold from 10*g* to 15*g*, the number of impacts was reduced by 45% and 81% of impacts were removed.[105] Therefore, studies with lower trigger values will record more impacts and may tend to show lower mean impact magnitudes than studies with higher trigger values.

Summary of Device Limitations and Implications for Interpretation

The descriptions of each device and their relative limitations should be considered when examining the data that we have summarized. Moreover, caution should be used when comparing results across studies in which different devices were examined. To facilitate the interpretation of data based on similar methods, data for helmeted devices are summarized in Table 2, and data for nonhelmeted devices are summarized in Table 3.

In addition, head-impact data, regardless of the sport or collection device, are heavily right skewed, with most impacts tending to have lower magnitudes. Therefore, means reported are greater than medians reported, indicating that the measure of central tendency should be considered when interpreting results. Whereas means and standard deviations are commonly provided, medians and interquartile ranges are better assessments of the typical impact magnitude.

Normative Data

Normative data for head-impact exposure and magnitudes by level, sport, and sex are summarized. We compared the recorded impacts of helmeted impact-monitoring systems (HITS and 6DOF) across age and sport in Table 2. We summarized data collected from nonhelmeted impact-monitoring devices (X2 X-Patch and X-Guard) in Table 3. Impacts per season, impacts per player per session, peak resultant LA, peak resultant AA, and HITsp were extracted from each article when available. Impact magnitudes are described by means, medians, and 95th percentiles.

Helmeted Designs. ***Football.*** Collegiate football athletes sustained 420 to 1177 impacts,[48,55] with 9.4 impacts per session[49] at a median linear magnitude of 19*g*[50] and average

Table 2. Summary Data for Helmeted Devices Extended on Next Page

Study[a]	Substudy[b]	Years	Sport	No. of Players	Total Impacts by Group	Sex	Age
Duhaime et al[16] (2012)		2007–2011	Football and ice hockey	450	486 594 Concussive = 31	Male and female	College
	Wilcox et al[42] (2014)	2009–2012	Ice hockey	41 Male[c]	19 880	Male and female	College
				58 Female	17 531		
	Wilcox et al[43] (2015)	NP	Ice hockey	58	Concussive = 4	Female	College
	Wilcox et al[44] (2014)	2009	Ice hockey	23 Male	1965	Male and female	College
		2008–2011	Ice hockey	31 Female	2532		
	Brainard et al[45] (2012)	2008	Ice hockey	37 Male	Total = 15 281	Male	College
				51 Female	Total = 12 897	Female	College
Rowson et al[46] (2014)		2005–2010	Football	1833	1 281 444	Male	College
	Funk et al[47] (2011)	2006–2010	Football	98	Subconcussive = 37 128 Concussive = 4	Male	College
	Crisco et al[48] (2011)	2007–2009	Football	314	Total = 286 636		College
	Crisco et al[49] (2010)	2007	NP	188	NP	Male	College
	Duma and Rowson[50] (2009)	2007–2008	Football	NP	6 Degrees of freedom = 4709	Male	College
		2003–2008	Football		HITsp = 71 300		
	Rowson et al[26] (2011)	2007–2009	Football	HITsp = 314	Subconcussive = 193 465	Male	College
					Concussive = 33		
				6 Degrees of freedom = 21	14 341		
	Rowson et al[31] (2009)	2007	Football	10	1712	Male	College
	Rowson et al[51] (2011)[d]	2006–2007	Football	3		Male	College
	Gwin et al[52] (2010)	2005–2006	Football	40	20 733	Male	College
	Brolinson et al[53] (2006)	2003–2004	Football	52	Total = 11 604	Male	College
	Duma et al[54] (2005)	2003	Football	38	Total = 3312	Male	College
Gysland et al[55] (2011)		NP	Football	46	NP	NP	NP
	Ocwieja et al[56] (2011)	2010	Football	46	Total = 7992 Special teams = 2250	Male	College
Liao et al[57] (2016)		2004–2011	Football	24	Concussive = 444	Male	College
	Guskiewicz et al[58] (2007)	NP	Football	88	Total = 104 714	Male	College

Table 2. Extended From Previous Page, Continued on Next Page

Minimum Recording	No. of Impacts Mean ± SD or Median (IQR)	Resultant Linear Acceleration Mean ± SD or Median (IQR)	Resultant Linear Acceleration 95th Percentile	Resultant Angular Acceleration Mean ± SD or Median (IQR)	Resultant Angular Acceleration 95th Percentile	HITsp Mean ± SD or Median (IQR)	HITsp 95th Percentile
NP	NP	NP	NP	NP	NP	NP	NP
	NP	86.1 ± 42.6	NP	3620 ± 2166	NP	NP	NP
NP	Per season = 287 (200–446)	15.7 (14.8–17.1)	41.6	1630 (1454–1733)	4424	13.6 (13.4–14.1)	NP
	Per season = 170 (116–230)	15.0 (14.5–15.5)	40.8	1211 (1091–1353)	3409	13.1 (12.9–13.6)	NP
NP	NP	43.0 ± 11.5 44.2 (34.7–52.5)	NP	4029.5 ± 1434.8 4557.0 (3615.5–4907.0)	NP	25.6 ± 4.8	NP
NP	NP	31.2 ± NA	NP	2881 ± NA	NP	18.8 ± NA	NP
NP	NP	28.3 ± NA	NP	1766.8 ± NA	NP	16.7 ± NA	NP
9.6	Per season = 347.3 ± 170.2 Per session = 2.9 ± 1.2	NP	NP	NP	NP	NP	NP
9.6	Per season = 179.2 ± 80.5 Per session = 1.7 ± 0.7	NP	NP	NP	NP	NP	NP
NP	NP	NP	NP	NP	NP	NP	NP
10.0	NP	NP	NP	NP	NP	NP	NP
	NP	145 ± 35	NP	NP	NP	NP	NP
NP	Per season = 420 (217–728)	NP	62.7	NP	4378	NP	32.6
14.4	Per session = 9.4 (NA)	NP	NP	NP	NP	NP	NP
NP	NP	17 (NA)	NP	931 (NA)	NP	NP	NP
	NP	19 (NA)	NP	NP	NP	NP	NP
14.4	NP	NP	NP	1230 ± 915 981 (NA)	NP	NP	NP
NP	NP	NP	NP	5022 ± NA 4948 (NA)	7688	NP	NP
NP	NP	NP	NP	1158 ± 972 872 (NA)	NP	NP	NP
10.0	NP	NP	NP	NP	NP	NP	NP
10.0	NP	17.5 (NA)	NP	1017 (NA)	NP	NP	NP
10	NP	NP	NP	NP	NP	NP	NP
10.0	NP	20.9 ± 18.7 15.3 (NA)	NP	NP	NP	NP	NP
10.0	NP	32 ± 25	NP	NP	NP	NP	NP
NP	Per season = 1177.3 ± 772.9	NP	NP	NP	NP	NP	NP
NP	NP	NP	68.5	NP	4960	NP	38.8
	NP	24.8 ± NA	NP	1430.4 ± NA	NP	15.6 ± NA	NP
NP	NP	97 (NA)	NP	5359.4 (NA)	NP	NP	NP
10.0	NP	NP	NP	NP	NP	NP	NP

Study[a]	Substudy[b]	Years	Sport	No. of Players	Total Impacts by Group	Sex	Age
					Concussive = 13		
	Mihalik et al[59] (2007)	2005–2006	Football	72	Total = 57 024	Male	College
	McCaffrey et al[60] (2007)	NP	Football	43	NP	Male	College
Ji et al[61] (2015)		2007–2011	Football and ice hockey	11	Concussive = 11	Male	High school and college
	McAllister et al[62] (2011)	NP	Football and ice hockey	10	Concussive = 10	Male	High school and college
Beckwith et al[17] (2013)		2005–2010	Football	Non-concussive days = 95	161 732	Male	High school and college
				Concussive days = 95			
	Beckwith et al[63] (2013)	2005–2010	Football	95 Concussed	161 732	Male	High school and college
Greenwald et al[40] (2008)		2004–2006	Football	259	289 916	Male	High school and college
Schnebel et al[64] (2007)		2005	Football	40	Collegiate = 54 151	Male	High school and college
				16	High school = 8326		
Schmidt et al[65] (2015)		NP	Football	49	Total = 19 775	Male	High school and college
	Schmidt et al[66] (2014)	NP	Football	37	16 066	Male	High school
Martini et al[67] (2013)		2009–2011	Football	Run-first offense = 42	Total = 22 091	Male	High school
				Pass-first offense = 41	Total = 13 527		
Broglio et al[68] (2011)		2007–2010	Football	95	Total = 30 298	Male	High school
	Eckner et al[69] (2011)	2007–2010	Football	NP	Concussive = 19	Male	High school
	Broglio et al[70] (2011)	2007–2010	Football	95	Concussive = 20	Male	High school
Broglio et al[71] (2013)		2009	Football	42	Total = 32 510	Male	High school
Broglio et al[72] (2010)		2005–2008	Football	78	Total = 54 247	Male	High school
	Broglio et al[73] (2009)	2007	Football	35	Total = 19 224	NP	NP
Urban et al[74] (2013)		NP	Football	40	NP	Male	High school
Davenport et al[75] (2014)		2012	Football	24	NP	NP	High school
Mihalik et al[76] (2011)		NP	Ice hockey	52	Total = 12 253	Male	13–16 y
	Mihalik et al[77] (2011)	2008	Ice hockey	37	Total = 7770	Male	13–16 y
	Mihalik et al[78] (2010)	NP	Ice hockey	16	NP	Male	15–16 y
	Mihalik et al[79] (2008)	NP	Ice hockey	14	Total = 4543	Male	13 y
Reed et al[80] (2010)		2006	Ice hockey	13	Total = 1821	Male	13–14 y
Daniel et al[81] (2014)		2012[e]	Football	10	Total = 2098	Male	12–14 y
Munce et al[82] (2015)		NP	Football	22	Total = 6183	Male	11–13 y

Table 2. Extended From Previous Page, Continued on Next Page

Minimum Recording	No. of Impacts Mean ± SD or Median (IQR)	Resultant Linear Acceleration Mean ± SD or Median (IQR)	Resultant Linear Acceleration 95th Percentile	Resultant Angular Acceleration Mean ± SD or Median (IQR)	Resultant Angular Acceleration 95th Percentile	HITsp Mean ± SD or Median (IQR)	HITsp 95th Percentile
	NP	102.8 ± 32.0 100.4 (84.1–109.9)	NP	5311 ± NA 5299 (NA)	NP	NP	NP
10.0	NP	22.3 ± 1.8	NP	NP	NP	NP	NP
NP	NP	NP	NP	NP	NP	NP	NP
NP	NP	69.7 ± 24.1 70.1 (57.8–83.2)	NP	4506 ± 1619.2 4718 (4007–5469.5)	NP	NP	NP
NP	NP	73.6 ± 21.3	NP	5025 ± 1226	NP	NP	NP
14.4	NP	NP	63.5	NP	2761	NP	NP
	NP	NP	82.0	NP	3376	NP	NP
14.4	NP	102.5 ± 33.8	NP	3977 ± 2272	NP	NP	NP
NP	NP	NP	NP	NP	NP	NP	NP
10.0	NP	NP	NP	NP	NP	NP	NP
	NP	NP	NP	NP	NP	NP	NP
NP	NP	NP	NP	NP	NP	NP	NP
NP	NP	NP	NP	NP	NP	NP	NP
15.0	Per season = 455.8 ± 192.6	25.7 ± 15.3	NP	1675.4 ± 1183.9	NP	15.5 ± 7.9	NP
	Per season = 303.7 ± 148.0	28.6 ± 17.8	NP	1777.6 ± 1266.6	NP	16.2 ± 9.3	NP
15.0	Per season = 652 ± NA and 626 (NA)	NP	NP	NP	NP	NP	30.5
NP	Per season = 704 ± NA	NP	55.5	NP	3901	NP	29.0
15.0	NP	93.6 ± 27.5	NP	6402.6 ± 1753.9	NP	63.4 ± 20.0	NP
14.4	Per season = 774 ± 502	25.9 ± 15.5	NP	1694 ± 1215.9	NP	15.6 ± 8.2	30.5
15.0	NP	25.1 ± 15.4	NP	1627.1 ± 1182.9	NP	NP	NP
15.0	Per session = 15.87 ± 17.87	NP	NP	NP	NP	NP	NP
14.4	Per season = 185 (NA)	21.9 (NA)	57.6	973 (NA)	2481	NP	NP
10.0	NP	NP	NP	NP	NP	NP	NP
10.0	Per season = 223 (NA) Per game = 61 (NP)	18.4 ± NA	45.6	1464.5 ± NA	4150	14.1 ± NA	26.8
10.0	NP	17.5 ± NA	NP	1587.7 ± NA	NP	14.0 ± NA	NP
10.0	NP	21.5 ± NA	NP	1441 ± NA	NP	15.8 ± NA	NP
10.0	NP	19.0 ± NA	NP	NP	NP	NP	NP
10.0	Per season = 140.1 ± 16.72 Per game = 5.19 ± 0.62	22.12 ± NA	NP	1557.4 ± NA	NP	NP	NP
NP	Per season = 210 ± 162[e]	26 ± 18 21 (NA)	61	1082 ± 846 898 (NA)	2571	NP	NP
NP	Per season = 252 (NA)	25.5 ± NA 20.2 (NA)	57.3	1691.8 ± NA 1407.4 (NA)	NP	NP	NP

Study[a]	Substudy[b]	Years	Sport	No. of Players	Total Impacts by Group	Sex	Age
Cobb et al[83] (2013)		NP	Football	50	Total = 11 978	Male	9–12 y
Young et al[84] (2014)		2013	Football	14	Total = 4039	Male	9–11 y
Daniel et al[85] (2012)[f]		NP	Football	6 Degrees of freedom = 7	748	Male	6–9 y
Young et al[86] (2014)[f]		2011–2012	Football	Head Impact Telemetry = 7	Total = 3059	Male	6–8 y
Hanlon and Bir[87] (2012)		NP	Soccer	24	Headers = 47 Nonheaders = 20	Female	<14 y
Dickson et al[88] (2016)		2009–2011	Snow sports	107	Total = 970	Male and female	9–18 y
Stojsih et al[89] (2008)		NP	Boxing	27 Males 38 Females	Total = 1128 Total = 802	Male and female	22–24 y

Abbreviations: HITsp, Head Impact Telemetry severity profile; IQR, interquartile range; NA, not available; NP, not provided.
[a] Indicates main study.
[b] Indicates studies from which main study compiled data.
[c] Subset participated in 2010 season only.
[d] Study used 6 degrees of freedom.
[e] Partial season.
[f] Study used 6 degrees of freedom and Head Impact Telemetry.

linear magnitudes ranging from 20*g* to 22*g*.[53,59] The 95th percentile for linear magnitudes was from 63*g* to 69*g*.[48,56] Rotational median magnitude has been reported[50] at 931 rad/s^2, and 95th percentiles ranged[48,56] from 4378 to 4960 rad/s^2. The 95th percentile for HITsp ranged[48,56] from 33 to 39. Greenwald et al[40] reported an HIC15 value of 67.9 for all impacts on the basis of mixed high school and collegiate datasets for football athletes.

High school football players averaged 652 to 774 impacts per season.[68,69,71,74] Athletes sustained an average of 24.1 impacts per game[71] or 15.87 impacts per game and practice combined.[73] Linear magnitudes averaged 26*g*,[71] with a median of 22*g*.[74] Angular magnitudes averaged[71,72] 1627 to 1694 rad/s^2 and had a median value[74] of 973 rad/s^2.

Youth football impact data have been reported for athletes ranging in age from 6 to 14 years.[81–84,86] Impacts per player per season ranged[81–84,86] from 161 to 345, with a dose-response relationship of younger athletes sustaining fewer impacts per season. Players between the ages of 9 and 12 years sustained 10.6 impacts per session.[83] A similar dose-response relationship emerged with age and impact magnitudes. Linear magnitudes had median values ranging from 16*g* to 21*g*, and angular median values ranged[81–84,86] from 686 to 1407 rad/s^2.

Player position also influenced the impact profile, given that linemen sustained the highest number of impacts but at relatively low magnitudes.[48,49,64,68,71] Defensive-line players sustained more impacts per session than offensive linemen, offensive-skill players, and defensive-skill players. Defensive-line players also had impacts of greater linear magnitudes than did defensive-skill and offensive-line players.[73] Offensive- and defensive-line players sustained greater average angular magnitudes than did offensive- and defensive-skill players.[73] Whereas quarterbacks sustained the fewest overall impacts, they had the greatest linear magnitudes, and tight ends, running backs, and linebackers had the greatest angular values.[48,71]

Ice Hockey. Collegiate ice hockey players sustained 170 to 347 impacts per season.[42,45] Mihalik et al[76] observed that high school ice hockey players had 223 impacts per season, with 61 per game and 22 per practice. Reported linear magnitudes ranged from 18*g* to 22*g* for high school players.[76–78] The 95th percentile was greater for high school (46*g*)[76] than for collegiate (42*g*) players.[42] High school male athletes had lower rotational accelerations, reportedly[77–79] averaging 1441 to 1588 rad/s^2 and a 95th percentile[76] of 4150 rad/s^2. The impact-severity measure HITsp was reported[76–78] to be 14 to 16, with a 95th percentile[76] of 26.8.

Among male and female collegiate ice hockey players, men sustained more head impacts in games due to contact with another player and the board than women did.[44] Overall, men had a greater mean number of impacts than women per athlete-exposure (2.9 and 1.7 impacts per athlete-exposure, respectively)[45] and per season (287.0 and 168.8 impacts per season, respectively).[42] Men sustained a

Table 2. Extended From Previous Page

Minimum Recording	No. of Impacts Mean ± SD or Median (IQR)	Resultant Linear Acceleration Mean ± SD or Median (IQR)	Resultant Linear Acceleration 95th Percentile	Resultant Angular Acceleration Mean ± SD or Median (IQR)	Resultant Angular Acceleration 95th Percentile	HITsp Mean ± SD or Median (IQR)	HITsp 95th Percentile
14.4	Per season = 240 ± 147	18 ± 2	43 (7)[c]	856 ± 135	2034	NP	NP
	Per session = 10.6 ± 5.2	19 (NA)	46	890 (NA)	2081		
10.0	Per season = 345 ± 165 Per game = 8 ± 5 Per practice = 16 ± 8	20 ± 2	44 (6)[c]	866 ± 118	2050 ± 366	NP	NP
NP	NP	18 ± NA 19 (NA)	40	901 ± NA 671 (NA)	2347	NP	NP
14.4	Per season = 161 ± 111	16 (NA)	38 ± 13	686 (NA)	2052	NP	NP
10.0	NP	NP	NP	NP	NP	NP	NP
NP	NP	19.6 ± 11.4 18.5 (11.8–23.9)	NP	1707.2 ± 1252.8 1247.1 (826–2407.1)	NP	NP	NP
10.0	Per session = 6.0	NP	NP	NP	NP	NP	NP
9.6	NP	30 ± 21	NP	2571 ± 1852	NP	NP	NP
NP	NP	28 ± 17	NP	2533 ± 1524	NP	NP	NP

greater median number of impacts per season (287 [interquartile range = 200–446]) than women (170 [interquartile range = 116–230]).[42] Although more impacts were sustained during games than during practices for both sexes, overall men sustained more impacts per practice and per game.[42,45]

The distribution of impact locations was similar between sexes for all locations except the left and right sides, where men (right = 13.6%, left = 13.8%) sustained a lower frequency than women (right = 15.4%, left = 14.9%).[45] Linear peak magnitudes to the front and side of the head were greater for men than for women, and angular peak magnitudes were greater in all 4 locations (front, side, top, and back) for men.[42,44] Using arbitrary cutoffs, men appeared to sustain more high-magnitude impacts than women, whereas women sustained more lower-magnitude impacts than men. Men were 1.3 times more likely to sustain impacts with a linear magnitude greater than 100*g* and 1.9 times more likely to sustain an angular impact greater than 5000 rad/s,[2] whereas women were 1.1 times more likely than men to sustain an impact less than 50*g*.[45]

Among ice hockey players, researchers reported no differences in the frequency of impacts[42,79] or magnitudes between offensive and defensive positions.[42,76,79] However, Reed et al[80] observed that wingers (10.92 ± 0.50) had a greater number of head impacts than centers (6.45 ± 0.72) or defensive players (5.95 ± 0.44). Moreover, wingers experienced greater angular-rotation magnitudes than centers, and defensive-position players had greater rotational values than centers.[80] Position did not play a role at the youth, high school, or collegiate levels for resultant LA.[42,76,79,80]

Soccer. In the only youth soccer study, Hanlon and Bir[87] investigated header and nonheader impacts in a youth female population using a modified HITS device. Nonheader impacts averaged 20*g* and had an angular average magnitude of 1247 rad/s^2.

Nonhelmeted Designs. ***Football.*** Whereas the HITS and 6DOF devices fit into a football helmet, some researchers[90–92] have used the X2 X-Patch to study head impacts among collegiate, high school, and youth athletes. Swartz et al[91] observed that collegiate football players sustained 13.8 ± 7.27 impacts per session. Full-pad practices were associated with the greatest linear (28.8*g*) and angular (5605 rad/s^2) magnitudes, followed by games (28.2*g* and 5560 rad/s^2, respectively) and helmet-only practices (21.7*g* and 3899 rad/s^2, respectively).[90] Compared with collegiate football players, youth football players (aged 8–15 years) sustained fewer average impacts: 12.9 ± 3.9 impacts per game and 7.5 ± 3.4 impacts per practice.[92]

Soccer. Soccer is a nonhelmeted sport requiring a different impact-monitor design. The X2 X-Patch has been used in collegiate, high school, and youth soccer to evaluate head impacts.[93–95] For high school and collegiate athletes, all soccer-related head impacts averaged 37.6*g* resultant LA and 7523 rad/s^2 resultant AA.[47] Collegiate female soccer players sustained an average of 4.6 impacts per session (almost 7 per game and 3.5 during practice)[94] and resultant AA averages ranged[93,94] from 5626 to 7713 rad/s^2. High school female soccer players sustained an average of 2 impacts per session: 2.9 per game and 1.7 during practice.[94] At the youth level, male and female soccer players experienced a median linear impact magnitude of 18.3*g*.[95]

Rugby. Authors of the 2 studies that evaluated head impacts in rugby used the X-Guard[96] or X2 X-Patch.[97] For their X-Patch study, King et al[97] studied youth athletes who sustained a median of 10 impacts per match. The median linear and angular magnitudes were 15*g* and 2296 rad/s^2, respectively.[97] Using the X-Guard among amateur rugby

Table 3. Summary Data for Nonhelmeted Devices

Study	Device[a]	Years	Sport	No. of Players	Total Impacts by Group	Sex	Age	Minimum Recording	No. of Impacts Mean ± SD or Median (IQR)	Resultant Linear Acceleration Mean ± SD or Median (IQR)	Resultant Linear Acceleration 95th Percentile	Resultant Angular Acceleration Mean ± SD or Median (IQR)	Resultant Angular Acceleration 95th Percentile
Reynolds et al[90] (2016)	X-Patch	2013	Football	20	Games	Male	College	10	NP	28.2 ± NA	NP	5560 ± NA	NP
					Full-pad practice				NP	28.8 ± NA	NP	5605 ± NA	NP
					Helmet-only practice				NP	21.7 ± NA	NP	3899 ± NA	NP
Swartz et al[91] (2015)	X-Patch	2014–2015	Football	25	Helmetless tackle intervention	Male	College	NP	Per session = 9.99 ± 6.10	NP	NP	NP	NP
				25	No intervention			NP	Per session = 13.84 ± 7.27	NP	NP	NP	NP
Kerr et al[92] (2015)	X-Patch	2014	Football	70	Total = 7478	Male	8–15 y	10	Per season = 106.8 ± 70.1	NP	NP	NP	NP
					Game = 4250			NP	Per game = 12.9 ± 3.9	NP	NP	NP	NP
					Practice = 3228			NP	Per practice = 7.5 ± 3.4	NP	NP	NP	NP
Press and Rowson[93] (2017)										25 ± 16	NP	5626 ± 4223	
	X-Patch	NP	Soccer	26	Total = 17 865	Female	College	10	NP				NP
					Games = 9350			NP	NP	32 ± 18	NP	7126 ± 4555	NP
					Practices = 8515			NP	NP	20 ± 13	NP	4541 ± 3597	NP
McCuen et al[94] (2015)	X-Patch	NP	Soccer	24	Overall	Female	College	NP	Per session = 4.59 ± NA	39.3 ± NA 31.2 (NA)	NP	7713 ± NA 6429 (NA)	NP
					Games			NP	NP	41.2 ± NA 31.9 (NA)	NP	8213 ± NA 6715 (NA)	NP
					Practices			NP	NP	37.7 ± NA 30.7 (NA)	NP	7297 ± NA 6301 (NA)	NP
				29	Overall		High school	NP	Per session = 2.05 ± NA	37.6 ± NA 30.5 (NA)	NP	7523 ± NA 6530 (NA)	NP
					Games			NP	NP	37.3 ± NA 30.9 (NA)	NP	8078 ± NA 6792 (NA)	NP
					Practices			NP	NP	37.7 ± NA 30.2 (NA)	NP	7095 ± NA 6371 (NA)	NP
Chrisman et al[95] (2016)	X-Patch	2014	Soccer	17	Total = 73	Male and female	<14 y	15	NP	18.3 (NA)	NP	NP	NP
King et al[96] (2015)	X-Guard	2013	Rugby	38	Overall = 20 687	Male	22 y	10	564 ± 618	22 ± 16.2	NP	3902.9 ± 3948.8	NP
					Match				95 ± 133	NP	NP	NP	NP
King et al[97] (2016)	X-Patch	2015	Rugby	14	Overall = 721	NP	<9 y	10	Per match = 10 (4–18)	15 (12–21)	46 (37–55)	2296 (1352–4152)	10 434 (9297–10 760)

Abbreviations: IQR, interquartile range; NA, not available; NP, not provided.

[a] X2 Biosystems, Inc, Seattle, WA.

players averaging 22 years of age, King et al[96] indicated that the average linear magnitude was $22g \pm 16.2g$ and average angular magnitude was 3902.9 ± 3948.8 rad/s^2.

Snow Sports. To examine head impacts among snow sports (eg, skiing, snowboarding), Dickson et al[88] modified a HITS device for snow-sport helmets. Amateur male and female skiers and snowboarders (age range = 9–18 years) experienced 6 impacts per session. More impacts were sustained during snowboarding sessions (9.8 per session, 1.9 per hour) compared with skiing sessions (4.0 per session, 1.1 per hour). Most impacts (61%) were less than $20g$, and 9% were greater than $40g$.[88]

Boxing. Other than ice hockey, boxing was the only sport for which researchers compared female and male athletes. On average, males (42 ± 27 impacts/boxer) sustained a greater number of impacts than females (29 ± 18 impacts/boxer).[89] Whereas average resultant LA and AA did not differ between sexes (Table 2), maximal peak magnitudes were greater for males than for females.[89]

Summary of Normative Data. The HITS data collected from the 2013 seasons to the present have a rotational correction applied to adjust the previous overestimation of resultant AA.[26] Data collected before the correction can be downloaded to apply the correction. Consequently, when comparing studies before and after 2013, it is important to consider whether the presence or absence of the resultant AA weight changes the interpretation of the data.

Most research has involved male athletes, predominantly in football (n = 40),* ice hockey (n = 12),[16,42–45,61,62,76–80] soccer (n = 4),[87,93–95] and boxing (n = 1).[89] Males sustained more impacts in ice hockey and boxing than females did. In boxing, male and female athletes received impacts with similar average magnitudes.

Player position influenced the frequency and magnitude of impacts. In football, linemen experienced more impacts of low magnitude. Skill-position players may experience less frequent impacts but at greater linear and angular magnitudes. In ice hockey, positions are more fluid than in football and may contribute to the mixed effect of position among ice hockey players.

Overall, age influenced the number, magnitude, and severity of impacts for football players. High school football players sustained 3 times more impacts than youth players, and collegiate players sustained up to 1.8 more impacts than high school athletes. We found it interesting that the relationship for age was minimized for impact magnitudes. Whereas younger age could be associated with lesser impacts, the degree of association was much less than for frequency of impacts. Given the large error rate associated with the X-Patch (see "Discussion"), comparisons among youth, high school, and collegiate soccer players should be cautious. The age relationship observed in football cannot be generalized to ice hockey and soccer due to limited data across age groups.

The difference in average resultant LA and AA between the 2 soccer studies can be attributed to different impact types and devices. Whereas Hanlon and Bir[87] limited their observations to heading impacts, McCuen et al[94] examined all direct and indirect head impacts. Thus, heading the ball produced lower-magnitude impacts,[87] averaging $20g$ and 1247 rad/s^2, than all soccer-related head impacts.[94] Moreover, McCuen et al[94] used the X-Patch, and Hanlon and Bir[87] used the HITS. Different recording thresholds and error rates also made it inappropriate to compare results between studies.

References 16, 17, 33, 40, 46–75, 81–83, 85, 86, 106.

Concussive Impacts

Of the 61 articles included in our review, 20 presented data on concussive events. A total of 304 concussive events were collected across collegiate football, collegiate male and female ice hockey, high school football, and youth ice hockey. Some concussions were described in multiple articles, yielding 227 unique concussive events (collegiate = 138, high school and collegiate = 45, high school = 25, youth = 6). We summarize the descriptive statistics of concussive events across age and sport. The average resultant LA and AA for concussive events across age and sport are plotted in Figure 2.

Concussive impacts in football were associated with linear magnitudes ranging from $69.7g$ to $145g$,[16,17,47,58,61,62,70,72] with a 95th percentile of 7688 rad/s^2.[26] Among football injuries, acceleration and impact location increased the specificity of injury prediction.[40,72] A total of 6 collegiate ice hockey concussions were captured using the HITS. Linear values ranged from $30.7g$ to $31.7g$ in an all-male sample[61,62] (n = 2) and $30.4g$ to $52.2g$ in an all-female sample[43] (n = 4). Researchers have reported average rotational values of 4030 rad/s^2 for females[107] and values ranging from 1307 to 5419 rad/s^2 for males.[43,61,62] The recorded linear magnitude of the only youth ice hockey concussion reported in the literature was $31.8g$, and the rotational value was 2911 rad/s^2. This impact represented the lowest recorded linear magnitude in the sample and was in the lowest 50% for recorded rotational accelerations.[78] A summary of average concussive magnitudes across sport and age is presented in Figure 2.

Using in vivo data, researchers have attempted to estimate concussion risk using resultant LA, resultant AA, location of impact, or a combination of all 3 variables. Resultant AA may be a better indicator of concussion risk than resultant LA; an impact greater than 6945 rad/s^2 had a 75% risk for concussion, whereas an impact of 7483 rad/s^2 had a 90% risk.[26] In their analysis comparing resultant LA, resultant AA, and combined resultant LA and AA for concussion prediction, Rowson and Duma[33] reported that resultant AA was the least predictive measure, resultant LA and combined resultant LA and AA were the most predictive indicators for a concussive event, and no difference was observed between the resultant LA and combined resultant LA and AA performance. In high school football players, Broglio et al[72] noted that peak resultant LA, resultant AA, and location of impact were key features in concussive injury. Setting injury-tolerance levels at $96.1g$ for resultant LA and 5582.3 rad/s^2 for resultant AA combined with impact location produced a high school athlete injury model with 21.3% specificity.[72]

Under certain circumstances, the HITsp metric that combines resultant LA, resultant AA, HIC15, and impact location estimated concussion among the top 1% and 2% of impacts more efficiently than resultant LA, resultant AA, or other impact-severity measures.[40] At the 75% positive predictive value, the false-positive rate was 0.78% for HITsp, 1.64% for resultant LA, 2.4% for resultant AA, and

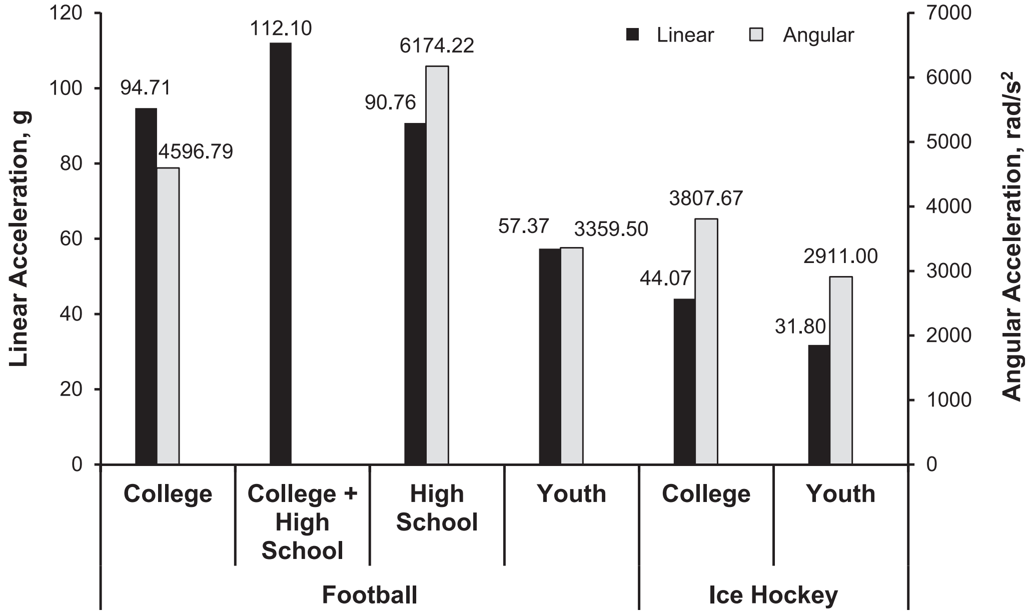

Figure 2. Linear and angular acceleration average values for concussive impacts: football and ice hockey. Individual concussive impact variables were extracted from multiple studies, and the means are presented. The number of concussive cases for each variable for football were collegiate linear (n = 82), collegiate angular (n = 105), high school and collegiate linear (n = 45), high school linear (n = 25), high school angular (n = 24), youth linear (n = 6), and youth angular (n = 4); and for ice hockey were collegiate linear (n = 6), collegiate angular (n = 6), youth linear (n = 1), and youth angular (n = 1).

1.87% for HIC15.[40] The HITsp metric produced 150 false-positive results (identifying nonconcussive events as concussive), equating to 6% specificity. The increased performance was not evident when examining all impacts or at the 90% positive predictive value.[40] Each of these models is limited by low specificity, indicating that most concussive impacts occur below these arbitrary threshold levels. For example, when assessing head impacts greater than 80*g* among collegiate players, researchers observed that concussions only occurred in 0.28% to 0.38% of impacts,[47,59,64] and nearly half of concussions will be undiagnosed.[47] Given that many concussions occur below the arbitrary 100*g* threshold and the risk for repeat concussion is 21% to 29% over 1 season and 33% to 50% over 5 seasons,[47] clinicians working on the sidelines should consider lowering the alert threshold for head-impact–monitoring devices. Funk et al[47] suggested using individualized impact thresholds so athletes can be evaluated each time they experience an impact more severe than a previous noninjurious impact. For example, if a player has sustained impacts ranging from 10*g* to 25*g* without a concussion, he or she will be evaluated for an impact larger than 25*g*. In addition, this implies that each evaluation will be tailored to an individual. If another player has had impacts up to 40*g* without injury, then he or she will only be evaluated when sustaining an impact greater than 40*g*. However, the caveat is that the number of false-positive alerts will increase, lowering the impact device's sensitivity and possibly overtaxing the clinical staff evaluating each athlete after each alert.

DISCUSSION

The primary purpose of our review was to describe the available head-impact–measurement technologies and summarize the concussion biomechanical research with a focus on their clinical utility for athletic trainers. Given that many researchers add participants to each subsequent analysis, the main objective was to organize and present comprehensive head-impact data across multiple age groups, sports, and devices, enabling us to draw overarching clinical conclusions based on the currently available literature.

Overall, age influenced impact frequency and magnitudes, with more impacts of greater magnitudes sustained with increasing age. Although player position in ice hockey had no consistent effect, linemen in football tended to sustain more impacts at lower magnitudes, but speed players and quarterbacks sustained impacts of greater magnitude. Player-position data were collected from high school and collegiate football athletes. It is unknown whether position plays a role in impact frequency and magnitude among youth players. Given that youth players are less likely to specialize in 1 position, it is reasonable to expect that position has no or a minimal effect on impact frequency and magnitudes among youth players. Sex affected impact frequency, with males sustaining more impacts in ice hockey and boxing. However, whereas males sustained greater impact magnitudes in ice hockey, magnitudes were similar across sexes in boxing.

Although no injury threshold exists, researchers have found criteria that increase the risk for concussion. A history of concussion, along with measures that combine resultant LA, resultant AA, and location of impact, influenced the likelihood of concussion.

Clinical Utility

The aim of assessing impact biomechanics is to associate the insult mechanism with a clinical outcome (eg, concussion). By understanding the injury mechanism, sports medicine professionals can make more informed decisions on improved equipment[46] and rule changes[91,92,107–109] to reduce concussion incidence.[8] Biomechanical data have effectively improved player safety via protective equipment and rule changes.[26,86,92,95,110–112] For example, the STAR helmet rating was initiated in 2011 at Virginia Tech to identify which helmets have a theoretical reduced concussion risk.[113] Combining laboratory and field data, the injury-risk function estimates the concussion risk for an individual player for 1 season.[26] Recently, the STAR rating expanded to include ice hockey helmets.[114]

In addition, the results of youth football studies[85] led the Pop Warner Little Scholars, Inc, youth football program to change to practice regulations that reduced head-impact exposure by 50%,[83] and high schools[115] and colleges[110] have begun to evaluate contact-practice regulations. Most notably, the Ivy League eliminated full-contact practices during the regular season starting in 2016.[110] By implementing a helmetless-tackling intervention during collegiate football practices, impacts were reduced by 27% per session.[91] The effectiveness of rule changes in reducing concussion incidence has been observed in youth ice hockey and football. Emery et al[107] compared body-checking with non–body-checking ice hockey leagues and found that 1 concussion per 1000 player hours could be averted by eliminating body checking. In youth football players aged 5 to 15 years, implementing the "Heads Up Football" coaching-education program reduced practice concussion rates from 0.79 concussions per 1000 athlete-exposures to 0.14 concussions per 1000 athlete-exposures, or 82%.[92] Whether equipment changes lead to improved player safety via reduced concussion incidence is unknown. More research is needed to determine whether mitigated athlete-exposure from equipment or rule changes reduces concussion incidences for all sports at all levels.

Head-impact sensors have limited applications to concussion diagnosis due to the error associated with individual impact measurements and other confounding factors that affect injury risk. However, the aggregate data that can be collected with each system have value. Given that single impacts have less effect on aggregate data, these systems that supply resultant LA, resultant AA, and impact counts can provide clinicians with estimates of player exposure. By understanding the strengths and limitations of each device, impact sensors may provide critical real-time data to monitor players. Concussion-risk estimates, however, are limited by athlete honesty and timely reporting of a concussion; up to 52% of concussions are unreported.[5,7] Moreover, an athlete's risk for concussion is influenced by myriad factors in addition to impact biomechanics. History of concussion, exposure, age, sex, migraine history, and comorbid psychiatric or learning disorders all influence concussion risk.[47,111,112,116,117] Viewing an athlete's head-impact data may provide context for the clinician working on the sidelines. However, impact sensors should not replace clinical judgment.

Other Commercially Available Devices

The Shockbox (i1 Biometrics, Kirkland, WA) is unlike either the HITS or X2 devices because it does not use traditional accelerometers. Rather, it uses binary force switches; 2 sensors measure front and rear impacts, and 2 other sensors measure side impacts.[118] Voltage activation of the force switches between 80 and 100 kHz is captured and calculated. At the moment of impact, the switch activates and reports the location, magnitude, and number of impacts.[119] Given that this device is not intended for research purposes, the threshold for the force switch to trigger is substantially higher: 30*g* in an in vivo study[119] and 50*g* in laboratory testing.[118]

Fourteen commercially available devices are advertised as head-impact–monitoring devices (Table 1), including helmeted (n = 6), nonhelmeted (n = 6), mixed (n = 1), and unknown designs (n = 1). No authors of peer-reviewed publications have validated or presented data collected from these devices. Whereas some devices capture acceleration magnitudes, others capture only the number of impacts in different risk zones. However, the normative values of these devices are unknown, thus providing little insight into the athlete's exposure. Given the unknown limitations of these devices and the minimal information they provide, clinicians and sideline staff should be skeptical of data output from these non–peer-reviewed devices. Ranging in price from $75 to $180, these devices are marketed to individual players. Therefore, clinical and sideline staff will likely be presented with impact data from 1 of the individually purchased devices. An overview of these devices and their metrics to help orient clinicians to the variety of devices they may observe and their various measurements is presented in Table 1.

Limitations

Clinical Utility. Perhaps most germane to clinical application, the head-acceleration devices that we discussed are currently the only way to measure head impacts in vivo. However, these devices have limitations, so sideline staff and clinicians should use caution when interpreting head-impact data. Given their limitations and the magnitude variations at which concussion occurs, these devices should not be used as the sole determinant for removing an athlete from participation. They cannot and should not replace the clinical expertise of trained medical staff.

Impact Measures. Head injury can be influenced by acceleration, change in momentum, and duration of impact. Currently reported peak acceleration values are only part of the equation. To gain a better understanding of injury mechanics, investigators should also examine the acceleration over time and force of impact. The HITS, X-Patch, and X-Guard produce acceleration-over-time plots, from which change in velocity (m/s^2) can be calculated. Change in velocity is the area under the acceleration-time plot. Force is not typically estimated because it would require knowing the effective mass of each person at the moment of impact. Therefore, acceleration is used because force is normalized by participant mass (acceleration = force / mass), allowing for between-subjects comparisons. Whereas force may appear useful in estimating injury risk, it would not discount the information obtained from resultant LA and AA. When a large defensive lineman tackles a smaller quarterback, one intuitively expects the quarterback to be propelled by the impact. The driver of this reaction is not force, because both players experience the same force at impact; however, given that the quarterback has less mass, he is accelerated more easily.

Currently, risk estimation for concussion has predominantly used acceleration variables to calculate risk; adding more biomechanical measures may improve the specificity of identifying concussive impacts.

Generalizability. Current biomechanical research on the head has largely focused on high school and collegiate athletes; therefore, the findings of these studies have limited generalizability to athletes younger than high school and to all females. Understanding concussive injury in younger athletes is important because children and adolescents may have prolonged recovery periods.[120] Moreover, female athletes sustain concussions at twice the rate of their male counterparts in some sports.[116] Researchers[121] have suggested this may result from more honest symptom reporting among females, the effect of estrogen, or decreased neck strength. However, no conclusive evidence has suggested that biomechanical differences in impact tolerance exist between sexes. Therefore, investigators should include more research on female athletes and less-represented sports, such as ice hockey, lacrosse, and soccer. Moreover, as sensors that can be used in nonhelmeted sports are developed, data can be collected in basketball, softball, field hockey, volleyball, and wrestling, which have greater than 3 concussions per 10 000 athlete-exposures.[122]

CONCLUSIONS

We described the current state of head-impact–monitoring research for concussion. Whereas many devices are available to collect head-impact data, most peer-reviewed devices are helmeted and have predominantly been used in football. The error rates and limited generalizability restrict the clinical utility of these devices. Although data collected in real time provide a measure of head-impact exposure, these devices should not be used for concussion diagnosis. However, by monitoring impact exposure, they have promoted design interventions that reduce the number of head impacts sustained by players over a season. Furthermore, proper interpretation of reported head-impact kinematics across age, sport, and position may inform future researchers and enable sideline staff clinicians to monitor athletes. Currently available head-impact–monitoring systems and algorithms have limited clinical utility due to error rates, design, and low specificity in predicting concussive injury.

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Address correspondence to Kathryn L. O'Connor, MS, NeuroTrauma Research Laboratory, University of Michigan, 401 Washtenaw Ave, Ann Arbor, MI 48109. Address e-mail to kloconn@umich.edu.