A review of impact testing methods for headgear in sports: Considerations for improved prevention of head injury through research and standards

T. Whyte[1,2], C.A. Stuart[1,2], A. Mallory[3,4], M. Ghajari[5], D.J. Plant[6], G.P. Siegmund[7,8], P.A. Cripton[1,2]

[1] Orthopaedic Injury Biomechanics Group, Departments of Mechanical Engineering and Orthopaedics, The University of British Columbia, Vancouver, BC, Canada;
[2] International Collaboration on Repair Discoveries (ICORD), The University of British Columbia, Vancouver, BC, Canada;
[3] Transportation Research Center Inc., OH, USA;
[4] The Department of Mechanical Engineering, Ohio State University, OH, USA;
[5] Dyson School of Design Engineering, Imperial College London, South Kensington Campus, SW7 2AZ, UK;
[6] Rheon Labs Ltd., 11S Hewlett House, Havelock Terrace, London, SW8 4AS, UK;.
[7] MEA Forensic Engineers & Scientists, 11-11151 Horseshoe Way, Richmond, BC V7A 4S5, Canada;
[8] School of Kinesiology, The University of British Columbia, Vancouver, BC, Canada;

Corresponding Author:
P.A. Cripton
Department of Mechanical Engineering
The University of British Columbia
Vancouver Campus
2054-6250 Applied Science Lane
Vancouver, BC, Canada, V6T 1Z4
cripton@mech.ubc.ca

Abstract

Standards for sports headgear were introduced as far back as the 1960s and many have remained substantially unchanged to present day. Since this time, headgear has virtually eliminated catastrophic head injuries such as skull fractures and changed the landscape of head injuries in sports. Mild traumatic brain injury (mTBI) is now a prevalent concern and the effectiveness of headgear in mitigating mTBI is inconclusive for most sports. Given that most current headgear standards are confined to attenuating linear head mechanics and recent brain injury studies have underscored the importance of angular mechanics in the genesis of mTBI, new or expanded standards are needed to foster headgear development and assess headgear performance that addresses all types of sport-related head and brain injuries. The aim of this review is to provide a basis for developing new sports headgear impact tests for standards by summarizing and critiquing: 1) impact testing procedures currently codified in published headgear standards for sports and 2) new or proposed headgear impact test procedures in published literature and/or relevant conferences. Research areas identified as needing further knowledge to support standards test development include defining sports-specific

head impact conditions, establishing injury and age appropriate headgear assessment criteria, and the development of headgear specific head and neck surrogates for at-risk populations.

Keywords: Headgear, helmet, impact testing, head injury, injury prevention, standards

1. Introduction

Protective headgear is used in many sports to reduce the risk of head injuries, being any insult to the scalp, skull or brain [1]. The introduction and development of sports headgear was typically in reaction to events of catastrophic head trauma such as death or severe brain injury [2,3]. Headgear has been remarkably successful in preventing these types of head injuries, with significant risk reductions realised in bicycling, American football (subsequently referred to as football), ice hockey (subsequently referred to as hockey), skiing and snowboarding [4–7]. This success is primarily related to preventing skull fractures, a role for which hard shell helmets have demonstrated effectiveness in literature dating back to the 1960s [8]. Skull protection, however, does not necessarily preclude all injury to the brain. The rate of sports-related mild traumatic brain injury (mTBI) has been steadily increasing over the past two decades, likely due to improvements in detection but possibly due to a true increase in incidence [9]. The evidence for headgear effectiveness in preventing mTBI is inconclusive in most sports [4,5,10–13], motivating new research efforts into evaluating headgear for reducing the risk of mTBI.

The most important aspect of sports headgear for prevention of head injury is how it performs in an impact, so long as the helmet is engaged in the impact (i.e. has adequate coverage), and remains in place for the impact events (i.e. has an effective retention system). Effective headgear prevents injury by redistributing the impact force on the head both spatially and temporally, thus attenuating the peak force and peak strain experienced by the tissues of the head. In order to assess impact performance of headgear designs, impact tests are performed in the laboratory. Ideally, headgear would be evaluated in all potential impact orientations, using a test surrogate that responds like a human and mimics injuries exactly as a human would sustain injuries, and assesses the risk of all possible head injuries. In reality, headgear is typically designed to satisfy the requirements of a limited number of impact tests defined in safety certification standards. These discrete tests are pragmatic simplifications of the wide range of impact events and human head impact responses that occur in sports.

The development of headgear standards for bicycling, football and hockey is well-documented. Impact tests for these applications first involved a guided drop of a humanoid headform and were introduced between 1961 and 1973. The standards built upon the then-existing helmet requirements for motorcyclists and motor sport participants [3,14–16], and considered the guided drop test representative of critical injurious impact scenarios occurring at the time. For instance, hockey helmet standard development was primarily concerned with minimising the force on the skull in a backwards fall and an occipital impact with the ice [15], Since the introduction of headgear standards and their success in preventing catastrophic head injuries, the landscape of head injuries in sports has changed. Today, the primary head injury of concern in hockey is mTBI resulting mainly from checking [17–19], an impact scenario considered to be of less importance during initial hockey helmet standards development [15]. This change in head injuries and injurious impact conditions has occurred in many sports, and warrants a review of certification standards and consideration of new requirements.

Injury biomechanics research has progressed considerably since the first headgear impact tests were devised and standards were implemented, particularly those aspects relating to brain injury. It is now known that many brain injuries, such as diffuse axonal injury and subdural haematoma, correlate with the rotational response of the head in an impact [20–27]. Since many current headgear impact testing standards neither measure nor allow rotation of the surrogate head, some researchers have asserted that these standards inadequately represent most real helmeted head impacts and thus have limited applicability for evaluating brain injury risk [2,28–36]. Many new impact test methods are emerging in headgear research, based on both recreating specific brain injury mechanisms and head impact scenarios that more closely resemble injurious events in sports. Novel headgear technologies are also being introduced, touting benefits in brain injury risk reduction [35,37–39]. In order to determine whether these promoted benefits will be realised in real-world impacts, it is important to consider and review the methods in which these novel headgear were tested. For future headgear standards, an appropriate method of assessing the risk, or the relative risk, of TBI is needed. The objective of this study is to provide a basis for developing new sports headgear impact tests for standards by summarizing and critiquing: 1) impact testing procedures in current headgear standards for sports and 2) new headgear test methods in published literature and/or at relevant conferences.

2. Impact attenuation testing in current headgear standards for sports

The impact performance requirements within headgear standards for sports, set by the Canadian Standards Association (CSA), the U.S. Consumer Product Safety Commission (CPSC), the American National Standards Institute (ANSI), the American Society for Testing and Materials (ASTM), the National Operating Committee on Standards for Athletic Equipment (NOCSAE), and the Snell Memorial Foundation, the European Committee for Standardization (CEN), and the International Organization for Standardization (ISO), are reviewed in Table 1. The following sections describe the impact condition, test surrogate and headgear assessment criteria specified for energy attenuation tests within the reviewed standards. Supplementary requirements for headgear within these standards, such as retention system requirements, are not reviewed.

Table 1 Summary of impact energy attenuation tests in headgear standards for sports

Headgear /helmet type	**Certification standards (last year of update)**	**Test apparatus**	**Impacts per site**	**Impact surface(s)[a]**	**Impact location**	**Impact severity[b]**			**Test failure threshold**
						Energy (J)	**Velocity (m/s)**	**Drop height (m)**	
Bicycling	CAN/CSA D113.2-M89 (2009) Cycle helmets[40]	ISO/DIS 6220 headform, magnesium or aluminium. Guided drop. Drop assembly mass 5 kg.	Single	Flat anvil Flat anvil Cylindrical anvil	Any point above test line	**80** **55** **55**	**5.7** **4.7** **4.7**	1.63 1.12 1.12	250 *g* 250 *g* 250 *g*
	CPSC 16 CFR Part 1203 (1998) Safety standard for bicycle helmets[41]	ISO/DIS 6220 headform, K1A magnesium. Guided drop. Drop assembly mass 5 kg.	Single	Flat anvil Hemispherical anvil Curbstone anvil	Any point above test line chosen to provide most severe test	96 58 58	**6.2** **4.8** **4.8**	**2.00** **1.20** **1.20**	300 *g* 300 *g* 300 *g*
	ANSI Z90.4 (withdrawn 1995) Protective headgear for bicyclists[42]	ISO/DIS 6220 headform. Guided drop. Drop assembly mass 5 kg.	Single	Flat anvil Hemispherical anvil	Any point above test line. On fasteners. At least one impact 12 mm above test line.	52 52	**4.57** **4.57**	**1.00** **1.00**	300 *g* 300 *g*
	Snell B-95 (1998) Standard for protective headgear for use with bicycles[43]	ISO/DIS 6220 headform, rigid low resonance metal. Guided drop. Drop assembly mass 5 kg.	Single	Flat anvil Hemispherical anvil Kerbstone anvil	Any point on or above the test line	**110** **72** **72**	6.63 5.37 5.37	2.24 1.47 1.47	300 *g* 300 *g* 300 *g*
	Snell B-90A (1998) Supplementary standard for protective headgear for use with bicycles[44]	ISO/DIS 6220 headform, rigid low resonance metal. Guided drop. Drop assembly mass 5 kg.	Single	Flat anvil Hemispherical anvil Kerbstone anvil	Any point on or above the test line	**100** **65** **58**	6.32 5.10 4.82	2.04 1.33 1.18	300 *g* 300 *g* 300 *g*
	Snell N-94 (1994) Standard for protective headgear for use in non-motorized sports[45]	ISO/DIS 6220 headform, rigid low resonance metal. Guided drop. Drop assembly mass between 5 and 6.5 kg.	Multiple (conditioning impacts prior to test impacts)	Flat anvil Hemispherical anvil Edge anvil Kerbstone anvil	Any point on or above the test line	**100** **72** **72** **72**	6.32 5.37 5.37 5.37	2.04 1.47 1.47 1.47	300 *g* 300 *g* 300 *g* 300 *g*
	ASTM F1447 (2012) Standard specification for helmets used in recreational bicycling or roller skating[46]	Rigid K1A-F ISO/DIS 6220, EN960 headform. Guided drop. Variable drop assembly mass.	Single	Flat anvil Hemispherical anvil Curbstone anvil	Any point on or above the test line	96 58 58	**6.2** **4.8** **4.8**	**2.00** **1.20** **1.20**	300 *g* 300 *g* 300 *g*

	ASTM F2032 (2015) Standard specification for helmets used in BMX cycling[47]	Rigid K1A-F ISO/DIS 6220, EN960 headform. Guided drop. Variable drop assembly mass.	Single	Flat anvil Hemispherical anvil Curbstone anvil	Any point on or above the test line	96 58 58	**6.2** **4.8** **4.8**	**2.00** **1.20** **1.20**	300 *g* 300 *g* 300 *g*
	ASTM F1952 (2015) Standard specification for helmets used for downhill mountain bicycle racing[48]	Rigid K1A-F ISO/DIS 6220, EN960 headform. Guided drop. Variable drop assembly mass.	Single	Flat anvil Hemispherical anvil Curbstone anvil	Any point on or above the test line	96 78 78	**6.2** **5.6** **5.6**	**2.00** **1.60** **1.60**	300 *g* 300 *g* 300 *g*
	EN 1078 (2014) Helmets for pedal cyclists and for users of skateboards and roller skates[49]	Metal EN 960 headform. Free-falling headform with a guided carrier. Variable mass headform.	Single	Flat anvil Kerbstone anvil	Impact sites within the test area to present worst-case conditions. On perceived weak areas.	73 52	**5.42** **4.57**	**1.50** **1.06**	250 *g* 250 *g*
Low speed scootering	ASTM F1898 (2015) Standard specification for helmets for non-motorized wheeled vehicle used by infants and toddlers[50]	Rigid K1A-F ISO/DIS 6220, EN960 headform. Guided drop. Variable drop assembly mass.	Single	Flat anvil Hemispherical anvil Curbstone anvil	Any point on or above the test line	96 58 58	**6.2** **4.8** **4.8**	**2.00** **1.20** **1.20**	300 *g* 300 *g* 300 *g*
Skateboarding, roller skating	ASTM F1447 (2012)[46]	See “Bicycling”							
	ASTM F1492 (2015) Standard specification for helmets used in skateboarding and trick roller skating[51]	Rigid K1A-F ISO/DIS 6220, EN960 headform. Guided drop. Variable drop assembly mass.	Multiple (three on one site with flat anvil)	Flat anvil Cylindrical anvil Triangular hazard anvil	Any point on or above the test line	52 52 52	**4.57** **4.57** **4.57**	**1.00** **1.00** **1.00**	300 *g*
	EN 1078 (2014)[49]	See “Bicycling”							
Ice hockey, ringette	CSA Z262.1 (2015) Ice hockey helmets[52]	Rigid CEN 960 headform magnesium. Guided drop. Variable drop assembly mass.	Three	MEP pad	Crown, front, front boss, rear, rear boss, side and a non-prescribed site above the test line	51	**4.5**	1.03	275 *g*
	NOCSAE ND030 (2016) Standard performance specification for newly manufactured ice hockey helmets[53]	Urethane NOCSAE headform. Guided drop. Variable drop assembly mass.	Three	MEP pad	Front, front boss, side, rear boss and a random location on or above the test line	30 60 75	**3.46** **4.88** **5.46**	0.61 1.22 1.52	SI 300* SI 1200* SI 1200*

	ASTM F1045 (2015) Standard performance specification for ice hockey helmets[54]	Rigid K1A magnesium EN960 headform. Guided drop. Variable drop assembly mass.	Multiple (depending on conditioning)	MEP pad	Crown, front, front boss, rear, rear boss, side and two non-prescribed sites on or above the test line.	51	**4.5**	1.03	300 *g*
	ISO 10256-2 (2016) Protective equipment for use in ice hockey – Head protection for skaters[55]	Metal EN 960 three quarters headform. Free-falling headform with a guided carrier. Variable mass headform.	Multiple (depending on conditioning)	MEP pad	Crown, front, front boss, side, rear, rear boss and two non-prescribed sites on or above the test line	51	**4.5**	1.03	275 *g*
		Sectioned EN 960 magnesium headform. Guided drop. Variable drop assembly mass.	Multiple (depending on conditioning)	MEP pad		51	**4.5**	1.03	275 *g*
Skiing, snowboarding	CSA Z263.1 (2014) Recreational alpine skiing and snowboarding helmets[56]	Rigid K-1A magnesium EN960 headform. Guided drop. Variable drop assembly mass.	Three	MEP pad	Any location at or above the test line	51	**4.5**	1.03	250 *g*
	Snell RS-98 (1998) Helmet standard for use in recreational skiing and snowboarding[57]	ISO/DIS 6220 headform, rigid low resonance metal. Guided drop. Drop assembly mass 5 kg.	Single	Flat anvil Hemispherical anvil Edge anvil	Any point on or above the test line	**100** **80** **80**	6.32 5.66 5.66	**2.0** **1.6** **1.6**	300 *g* 300 *g* 300 *g*
	Snell S-98 (1998) Helmet standard for use in skiing[58]	ISO/DIS 6220 headform, rigid low resonance metal. Guided drop. Drop assembly mass between 5 and 6.5 kg.	Single	Flat anvil Hemispherical anvil Edge anvil	Any point on or above the test line	**120** **100** **100**	6.93 6.32 6.32	**2.4** **2.2** **2.2**	300 *g* 300 *g* 300 *g*
	ASTM F2040 (2011) Standard specification for helmets used for recreational snow sports[59]	Rigid K1A-F ISO/DIS 6220, EN960 headform. Guided drop. Drop assembly mass 5 kg.	Single	Flat anvil Hemispherical anvil Edge anvil	Anywhere on or above the test line	96 58 58	**6.2** **4.8** **4.8**	**2.0** **1.2** **1.2**	300 *g* 300 *g* 300 *g*

	EN 1077 (2007) Helmets for alpine skiers and snowboarders[60]	Metal EN 960 headform. Free or guided fall of the headform. Variable mass headform.	Single	Flat anvil	Within the defined test area	73	**5.42**	**1.5**	250 *g*
Short track speed skating	ASTM F1849 (2012) Standard specification for helmets used in short track speed ice skating (not to include hockey)[61]	Rigid K1A-F ISO/DIS 6220, EN960 headform. Guided drop. Drop assembly mass 5 kg.	Single	Flat anvil Skate blade anvil	Any point on or above the test line	96 36	**6.2** **3.8**	**2.0** **0.75**	300 *g* 300 *g*
Lacrosse	NOCSAE ND041 (2015) Standard performance specification for newly manufactured lacrosse helmets with faceguard[62]	Urethane NOCSAE headform. Guided drop. Variable drop assembly mass.	Three	MEP pad	Front, front boss, side, rear boss and a random location on or above the test line	30 60 75	**3.46** **4.88** **5.46**	0.61 1.22 1.52	SI 300* SI 1200* SI 1200*
Football	NOCSAE ND002 (2015) Standard performance specification for newly manufactured football helmets[63]	Urethane NOCSAE headform. Guided drop. Variable drop assembly mass.	Multiple (up to five)	MEP pad	Front, front boss, side, rear boss and a random location on or above the test line	30 45 60 75 75	**3.46** **4.23** **4.88** **5.46** **5.46**	0.61 0.91 1.22 1.52 1.52	SI 300* SI 1200* SI 1200* SI 1200* SI 1200*
	ASTM F717 (withdrawn 2017) Standard specification for football helmets[64]	Rigid ISO/DIS 6220 headform. Guided drop. Drop assembly mass 5 kg.	Multiple (up to six)	MEP pad	Front, front boss, side, rear boss, rear and crown.	75 75	**5.47** **5.47**	1.53 1.53	275 *g* 300 *g*
Baseball, softball, T-ball	NOCSAE ND022 (2015) Standard performance specification for newly manufactured baseball/softball batter's helmets[65]	Urethane NOCSAE headform. Variable headform mass. Fired projectile.	Single	Softball Baseball	Front, front boss, side, rear boss and a random location on or above the test line	53 54	**24.6** **26.8**	- -	SI 1200* SI 1200*
	NOCSAE ND024 (2017) Standard performance specification for newly manufactured baseball/softball	Urethane NOCSAE headform. Guided drop. Variable drop assembly mass.	Two	Steel half cylinder anvil	Front, front boss, side, rear boss and a random location on or above the test line	45 45	**4.23** **4.23**	0.91 0.91	SI 1200* SI 1200*

	catcher's helmets with faceguard[66]	Urethane NOCSAE headform. Variable headform mass. Fired projectile.	Single	Softball Baseball		53 54	**24.6** **26.8**	- -	SI 1200* SI 1200*
	NOCSAE ND029 (2015) Standard performance specification for newly manufactured baseball/softball fielder's headgear[67]	Urethane NOCSAE headform. Variable headform mass. Fired projectile.	Single	Softball Baseball	Front, front boss, side, rear boss and a random location on or above the test line	63 65	**27.0** **29.5**	- -	SI 1200* SI 1200*
Equestrian, horse racing, harness racing, polo	Snell E2016 (2016) Standard for protective headgear for use in horseback riding[68]	ISO/DIS 6220 headform, rigid low resonance metal. Guided drop. Variable drop assembly mass.	Single	Flat anvil Hemispherical anvil Horseshoe anvil	Any point on or above the test line	92 74 64	**6.06**** **5.42**** **5.07****	1.87 1.50 1.31	275 *g*** 275 *g*** 275 *g***
	Snell H2000 (2000) Helmet standard for use in harness racing[69]	Rigid DOT FMVSS 218 low resonance metal headform. Guided drop. Drop assembly mass between 5 and 6.5 kg.	Single	Flat anvil Hemispherical anvil Horseshoe anvil	Any point on or above the test line	**145** **145** **145**	7.62 7.62 7.62	**2.9** **2.9** **2.9**	300 *g* 300 *g* 300 *g*
	ASTM F1163 (2015) Standard specification for protective headgear used in horse sports and horseback riding[70]	Rigid K1A-F ISO/DIS 6220, EN960 headform. Guided drop. Variable drop assembly mass.	Single	Flat anvil Equestrian hazard anvil	Anywhere above the centre of the impact line. At least one at the front, rear, or side.	90 63	**6.0** **5.0**	**1.8** **1.3**	300 *g* 300 *g*
	NOCSAE ND050 (2015) Standard performance specification for newly manufactured polo helmets[71]	Urethane NOCSAE headform. Guided drop. Variable drop assembly mass.	Single on anvils Three on MEP pad	Hemispherical anvil Equestrian hazard anvil MEP pad	Front, front boss, side, rear boss and a random location on or above the test line	75 75 30 75 75	**5.46** **5.46** **3.46** **5.46** **5.46**	1.52 1.52 0.61 1.52 1.52	SI 1200* SI 1200* SI 300* SI 1200* SI 1200*
Bull riding	ASTM F2530 (2013) Standard specification for protective headgear with faceguard used in bull riding[72]	Rigid K1A-F ISO/DIS 6220, EN960 headform. Guided drop. Drop assembly mass 5 kg.	Single	Flat anvil Equestrian hazard anvil	Any point on or above the test line	90 63	**6.0** **5.0**	**1.8** **1.3**	300 *g* 300 *g*

Pole vaulting	ASTM F2400 (2016) Standard specification for helmets used in pole vaulting[73]	Rigid K1A-F ISO/DIS 6220, EN960 headform. Guided drop. Variable drop assembly mass.	Single	Flat anvil Curbstone anvil	Any point on or above the test line	96 58	**6.2** **4.8**	**2.0** **1.2**	300 *g* 300 *g*
Martial arts	ASTM F2397 (2015) Standard specification for protective headgear used in martial arts[74]	Rigid K1A-F ISO/DIS 6220, EN960 headform. Guided drop. Drop assembly mass 5 kg.	Multiple (up to four)	Flat anvil	Anywhere on or between the test lines and should represent the sites with the greatest risk of failure.	20 40	**3.0** **4.0**	0.46 0.82	100 *g* 300 *g*
		Aluminium covered with vinyl skin Hybrid III headform with segmented rubber and aluminium Hybrid III neck with centre cable hanging head-down from 25 kg steel mass. Spring loaded strike from aluminium tube.	Multiple (up to four)	Aluminium tube striker		56 144	**5.0** **8.0**	- -	50 *g* 150 *g*
Soccer	ASTM F2439 (2016) Standard specification for headgear used in soccer[75]	Aluminium covered with vinyl skin Hybrid III headform with segmented rubber and aluminium Hybrid III neck with centre cable. Guided drop. Drop assembly mass 8.8 kg.	Multiple (up to six)	Headform anvil Steel post anvil MEP pad	Anywhere within the impact area	64 18 28	**3.8** **2.0** **2.5**	0.74 0.20 0.32	80 *g* 80 *g* 92 *g*
Motorized sports	ASTM F3103 (2014) Standard specification for testing off-road motorcycle and ATV helmets[76] (Impact tests specified in FMVSS 218[77] at 10.1)	Rigid K-1A DOT FMVSS 218 headform. Guided drop. Variable drop assembly mass.	Two	Flat anvil Hemispherical anvil	Any point on the area above the test line	90 68	**6.0** **5.2**	**1.83** **1.38**	400 *g*; or more than 2 ms above 200 *g*; or more than 4 ms above 150 *g*

	Snell M2015 (2015) Standard for protective headgear for use with motorcycles and other motorized vehicles[78]	Rigid ISO/DIS 6220 low resonance metal headform. Guided drop. Variable drop assembly mass.	Two for flat and hemispherical anvils. One for edge anvil.	Flat anvil Hemispherical anvil Edge anvil	Any point on or above the test line	150 115	**7.75**** **6.78****	3.06 2.34	275 *g* 275 *g*
	Snell SA2015 (2015) Standard for protective headgear for use in competitive automotive sports[79]	Rigid ISO/DIS 6220 low resonance metal headform. Guided drop. Variable drop assembly mass.	Two for flat and hemispherical anvils. Three for roll bar anvil. One for edge and kerbstone anvils.	Flat anvil Hemispherical anvil Roll bar anvil Kerbstone anvil Edge anvil	Any point on or above the test line. Specific lateral impact tests onto kerbstone anvil.	181 100 90 63 141	**8.50**** **6.31**** **6.00**** **5.00**** **7.50****	3.68 2.03 1.83 1.27 2.87	300 *g* 300 *g* 300 *g* 200 *g* 200 *g*
	Snell EA2016 (2016) Standard for protective headgear for use in elite automotive sports[80]	Rigid ISO/DIS 6220 low resonance metal headform. Guided drop. Variable drop assembly mass.	Two for flat and hemispherical anvils. Three for roll bar anvil. One for edge anvil.	Flat anvil Hemispherical anvil Roll bar anvil Edge anvil	Any point on or above the test line	219 90 90	**9.35**** **6.00**** **6.00****	4.46 1.83 1.83	275 *g* 275 *g* 275 *g*
	Snell K2015 (2015) Standard for protective headgear for use in kart racing[81]	Tested as per Snell SA2015 or Snell M2015.							
	Snell CM2016 (2016) Standard for protective headgear for use in children's motor sport activities[82]	Rigid ISO/DIS 6220 low resonance metal headform. Guided drop. Variable drop assembly mass.	Two for flat and hemispherical anvils. One for edge anvil	Flat anvil Hemispherical anvil Edge anvil	Any point on or above the test line. Chin bar.	150 90 76	**7.75** **6.00** **5.50**	3.06 1.83 1.54	290 *g* 290 *g* 275 *g*

[a] Anvils are rigid. MEP (Modular Elastomer Programmer) pad is a molded polyurethane thermoplastic elastomer with consistent and repeatable impact properties. Softball mass 166-174 g, circumference 10.875-11.125 inches (276.2 – 282.6 mm) and Compression-Displacement at 0.25 inches (6.4 mm) of no less than 300 lbs (1334.5 N). Baseball mass 142-149 g, circumference 9-9.25 inches (228.6 – 235.0 mm), Coefficient of Restitution of 0.5-0.55, Compression-Displacement at 0.25 inches (6.4 mm) of 200-300 lbs (889.6 – 1334.5 N) and construction as specified by Major League Baseball. Aluminium tube striker is 500 mm long from pivot point to strike point and a further 50 mm long beyond the strike point, external diameter of 80 mm and mass of 4.5 kg uniformly distributed along its length.

[b] Bolded values are specified in the standard. Values in grey have been calculated here based on the values provided in the standard. All calculations of impact energy, impact velocity and drop height are based on a drop assembly mass of 5 kg unless otherwise specified.

* SI (Severity Index). SI of 1200 has been approximated to be equivalent to 215g in a typical helmet impact.[83]

** Impact speed and peak allowable acceleration criterion change based on test headform size. Impact speeds and thresholds listed here are for headform size J.

2.1 Impact condition

In the reviewed headgear standards, the impact test methods typically involve a linear guided drop of the helmeted headform onto an impact surface as shown in Figure 1. This test method restricts the headform from rotating, limiting the motion to a single axis.

Another type of falling headform impact test, in addition to the guided headform drop, is specified in the ISO standard for hockey [55] and the CEN standard for alpine skiers and snowboarders [60]. The additional method involves a free-falling headform with a guiding carrier, shown in Fig. 2, which allows unrestricted headform motion following the impact event. This free-falling headform impact test is the only test method specified in the CEN standard for bicycle helmets [49].

Baseball and softball helmet standards do not use a falling headform drop test method and are instead tested by firing baseball and softball projectiles at a helmeted headform [65–67].

In addition to the guided headform drop test, martial arts headgear is tested in ASTM F2397-09 using an apparatus designed to strike the protected headform with an aluminum tube (see Fig. S1) [74].

Impact tests within martial arts and soccer headgear standards perform impact tests with the headform attached to the Hybrid III anthropometric test device (ATD) surrogate neck [74,75]. No other reviewed headgear standards use a flexible neck.

Predominant impact surfaces in headform drop tests are rigid anvils of various shapes (flat, hemispherical, edge, curbstone), or a Modular Elastomer Programmer (MEP) pad intended to represent a ground surface on which sports participants fall [75]. A number of headgear standards have anvils representative of hazards encountered in the specific sport. For example, a skate blade anvil is used in the speed skating helmet standard [61], a steel half cylinder anvil (representing a baseball bat) is specified in the baseball catchers' helmet standard [66], and a headform anvil (representing another player) and steel post anvil (representing a goal post) are referenced in the soccer headgear standard [75].

Impact severities (here characterized by the energy of the impact) range from 18 J (soccer headgear to steel post anvil [75]) to 145 J (harness racing helmet onto rigid anvils [69]). For reference, the impact energy specified in bicycle helmet standards ranges from 52 J to 110 J. The process for setting the impact energy for headgear testing in the reviewed standards is not clear. However, headgear standards with the highest impact energy drop tests appear to be the sports in which participants achieve the highest travelling speeds or highest falling distances, such as motorized sports (219 J), harness racing (145 J), skiing and snowboarding (120 J) and bicycling (110 J).

There was also variability within the reviewed standards regarding the number of impacts performed at one site on the headgear, likely related to the risk of repeated head impacts in each sport. Football, hockey, martial arts and soccer headgear are always subjected to multiple impacts per site whereas bicycle and equestrian helmets are typically subjected to one impact per site.

2.2 Human head test surrogate

Rigid metal headforms (such as a low-resonance magnesium K1A alloy headform), specified in ISO Draft International Standard (DIS) 6220 [84], European Standard EN 960 [85] or U.S. Department of Transport (DOT) Federal Motor Vehicle Safety Standard (FMVSS) 218 [77], are required for impact testing in many of the reviewed standards, see Table 1. Other headforms specified in the reviewed standards include the NOCSAE urethane headform [86] (Southern Impact Research Centre, LLC, Rockford, TN, USA) and Hybrid III ATD headform [87] (Humanetics, Huron, OH, USA).

Variable headform sizes are required for impact testing in most reviewed standards. For example, ASTM standards reference six rigid headform sizes and NOCSAE standards reference three urethane headform sizes [86,88]. Headgear is typically tested on the appropriate test headform size yet this is not always possible. NOCSAE standards prescribe shimming for helmets too large for the largest headform size so that the impacted area fits the head as intended if the helmet were a proper fit to the headform. Helmets too small for the smallest headform are not tested in NOCSAE standards but approved so long as the other sizes of that helmet model meet all requirements [86]. Martial arts headgear is tested with an appropriate size Hybrid III headform in ASTM F2397 [74]. Soccer headgear of all sizes is tested using the 50th percentile male Hybrid III headform in ASTM F2439 [75].

2.3 Assessment criteria and thresholds

The assessment criteria in the reviewed headgear standards relate to the linear acceleration response at the test headform centre of gravity during the impact. A peak linear acceleration criterion is the most common performance requirement, while NOCSAE standards use the Severity Index which involves integrating the linear acceleration over the time duration of the impact [89].

The allowable threshold of peak linear acceleration, expressed in *g* (standard acceleration due to gravity at 9.8 m/s^2), for sports headgear varies between sports as well as between standards for the same sport but typically lie in the range of 250-300 *g*. The aluminum tube striker test for martial arts headgear and the soccer headgear drop tests have lower threshold values of 80-150 *g* [74]. NOCSAE standards typically enforce a threshold limit of 1200 for the Severity Index, reportedly equivalent to 215 *g* “in a typical helmet impact” [83].

A number of headgear standards have separate performance requirements for a low and high severity impact. NOCSAE standards for hockey, lacrosse, football and polo include a Severity Index limit of 300 for a low velocity impact in addition to higher velocity impact tests with a Severity Index limit of 1200. The ASTM martial arts headgear standard specified a peak linear acceleration threshold of 100 *g* for a 20 J drop test and a 300 *g* threshold for a 40 J drop test [74].

3. New headgear test methods

Researchers have devised different test procedures for studying headgear impact performance beyond impact tests specified in standards. New (relative to impact tests in standards) headgear impact test methods in published literature and/or at relevant conferences are summarised in Table 2. This review categorised new impact test methods based on the type of headform, the impact opponent, the dynamics of the impact and the assessment criteria. Impact test severities achievable by these methods were variable (typically on a continuous scale) so tested impact speed is not included in Table 2. The review was not restricted to the types of sports headgear reviewed in Table 1 and includes impact mitigation studies that tested motorcycle helmets, industrial helmets, fall protection headgear and helmet accessories. Tests of these other headgear and accessories use biomechanical principles similar to those needed for sports impacts, hence their inclusion. Test methods relating to military helmets were excluded since the primary purpose of military headgear is protection from ballistic threats rather than from blunt impacts like those occurring in sports [90]. The following sections summarise and synthesize the main components of the new headgear test methods.

196 **Table 2 Review of new headgear test methods found in literature and/or at relevant conferences**

First author (year)	Headgear tested	Headform[c]	Head boundary condition at impact[c]	Pre-impact headform motion	Impact opponent/surface	Impact dynamics	Assessment criteria[d]	Image/diagram of test setup
Harrison (1996) [91] Halldin (2001) [35] Aare (2003) [28]	Motorcycle	Ogle (aluminium headform coated with PVC plastisol). Hybrid III.	None	Free-fall	Horizontally moving steel plate covered with 80 SiC grit grinding paper. Horizontally moving steel plate covered with polystyrene foam.	Free-fall of headform to impact horizontally moving impact surface	Linear and angular accelerations. Angular velocity change. Duration of angular acceleration.	Figure S2
Mills (2008) [92]	Bicycle	Ogle aluminium headform with PVC plastisol skin and acrylic wig.	None	Free-fall	Horizontally moving aluminium surface. Horizontally moving aluminium surface covered with 120-grade SiC grit grinding paper.	Headform free-falls onto horizontally moving impact surface	Linear and angular accelerations. Impact forces.	Figure S2
Chinn (2001) [93] Ghajari (2013) [94]	Motorcycle	Hybrid II	None	Free-fall	Various impact surfaces based on specific crash cases, e.g. textured slabs, hemisphere, kerbstone, bar, steel edge, vehicle components. Various impact angles from 15º to 90º.	Headform free-falls onto impact opponent	Linear and angular accelerations. Impact force.	Figure S3
Willinger (2014) [95] (2015) [96] Bourdet (2016) [97]	Proposals for bicycle and motorcycle helmets	Hybrid III	None	Free-fall	45º angled impact anvil covered with friction paper (80 gr).	Headform free-falls in different starting orientations onto the angled impact surface	Linear and angular accelerations. FE model criteria.	Figure S4
Bland (2018) [98]	Bicycle	Medium NOCSAE	None	Free-fall	45º angled steel anvil coated with 80 grit sandpaper.	Heaadform free-falls in different starting orientations onto the angled impact surface.	Linear acceleration. Change in angular velocity.	Figure S4

Ebrahimi (2015) [99]	Motorcycle	Size J rigid ISO/DIS 6220 EN 960	None	Free-fall	Flat steel anvil inclined at 15º or 30º to the vertical covered with grade 80 closed-coat aluminium oxide abrasive paper.	Headform free-falls in different starting orientations onto the angled impact surface	Linear and angular accelerations	Figure S4
Finan (2008) [36]	Football	Hybrid III	None	Free-fall	Flat anvil angled between -45º and 45º at 15º increments.	Headform free-falls onto the angled impact surface	Linear and angular accelerations. HIC.	Figure S4
Klug (2015) [100]	Bicycle	Hybrid III with modified mass and moment of inertia	None	Free-fall	Rigid anvil angled 30º from horizontal covered with 80 grain abrasive paper.	Headform free-falls onto angled impact surface	Linear and angular accelerations. Angular velocity. HIC36 HIP BrIC	Figure S5
Stigson (2017) [101]	Bicycle	Hybrid III	None	Free-fall	Rigid anvil angled 45º covered with grinding paper Bosch quality 40.	Headform free-falls onto angled impact surface	Linear and angular accelerations. Angular velocity. FE model criteria	Figure S6
Stuart (2017) [102]	Ski and snowsport	Hybrid III	None	Propelled at a variable angle toward the impact surface	Concrete and linoleum floor. Proposed for in situ snow surfaces.	Free-flight headform propelled using a spring-loaded carriage onto impact surface	Linear and angular accelerations	Figure S7
Saczalski (2016) [103]	Football	Hybrid III 5th percentile female	Hybrid III 5th percentile female neck	Swinging pendulum	Non-yielding surface.	Inverted headform and neck is swung using a “free pendulum” horizontally into the impact opponent.	Linear and angular accelerations. Angular velocity. Pulse duration. HIC.	Figure S8
Roseveare (2016) [39]	Motorcycle	Hybrid III	Hybrid III neck attached to a pendulum acting as a cantilevered mass	None	Rigid impact surface angled at 30º from vertical.	Angled impact surface falls in a guided-drop to impact the headform attached to cantilevered mass	Linear and angular accelerations. FE model criteria.	Figure S9
Dressler (2012) [104]	Ski	Hybrid III	Hybrid III neck attached to 16 kg drop carriage	Guided free-fall	Soft snow. Hard snow.	Guided drop of headform onto flat impact surface	Linear acceleration. HIC-15. Neck loads.	Figure S10

Dressler (2014) [105] Dressler (2018) [106]	Football	Hybrid III	Custom surrogate neck attached to 16 kg drop carriage	Guided free-fall	Steel impact platen covered with two layers of yoga mat foam.	Guided drop of headform onto impact surface	Linear and angular acceleration. HIC-15	Figure S11
Caccese (2014) [34] Ferguson (2015) [107] Seidi (2015) [108]	Protective headgear for falls	Hybrid III	Hybrid III neck attached to 8.2 kg twin-wire drop carriage	Guided free-fall	Concrete with vinyl tile. Steel plate. MEP pad.	Guided drop of headform onto flat impact surface	Linear and angular accelerations. HIC-15. RIC-36. PRHIC.	Figure S12
Pellman (2003) [109] Viano (2005) [110] (2006) [111] Withnall (2005) [112] Newman (2005) [113]	Soccer. Football.	Hybrid III	Hybrid III neck attached to drop rail	Guided free-fall	Hybrid III head and upper torso suspended from a hoist assembly. Artificial turf placed on rigid plate.	Guided drop of headform onto impact opponent	Linear and angular accelerations	Figure S13
Butz (2015) [114] Knowles (2017) [115]	Bicycle. Hockey. Helmet accessories. Wearable sensors.	Hybrid III	Hybrid III neck attached to gimbal (total mass 10 kg) on drop rail	Guided free-fall	Steel impact surface. MEP pad.	Guided drop of headform onto impact opponent	Linear and angular accelerations. Angular velocity. Impact forces. FE model criteria.	Figure S14
Clark (2015) [116] (2016) [117]	Hockey	Hybrid III	Unbiased neckform on drop rail	Guided free-fall	MEP pad.	Guided drop of headform onto flat impact surface	Linear and angular accelerations. Impact duration. FE model criteria.	Figure S15
Karton (2012) [118]	Speed skating	Hybrid III	Hybrid III neck attached to drop rail	Guided free-fall	Hemispherical nylon pad covering a MEP 60 Shore Type A disc secured to large metal anvil.	Guided fall of headform onto impact surface	Linear and angular accelerations	Figure S16
Posey (2006) [119]	Full90 soccer headguard	Hybrid III	Motorcycle ATD neckform on drop rail	Guided free-fall	Hybrid III head. Cylinder anvil.	Guided drop of headform onto impact opponent	Linear and angular accelerations. GAMBIT.	Figure S17

Aldman (1976) [120] (1978) [121]	Motorcycle	Ogle-Opat	Ogle-Opat neck attached to a drop rail carriage by a hinge upon a horizontal axis	Guided free-fall	Rotating ring of road surfaces: clean wood particle board (smooth), coarse grinding cloth (Naxos CKRG 20), angular stones, subangular-round stones, subangular-round stones with a thin layer of oil.	Guided drop of headform onto the rotating road surface	Linear and angular accelerations	Figure S18
Mills (1996) [30]	Motorcycle helmet materials (foam and shell)	2D slice with 12 mm thick web and 25 mm wide aluminium rim, outer diameter of 222 mm and mass of 2.255 kg	Attached with a rigid bar and pivots at the centre of the headform slice to a drop carriage which is attached to a drop rail	Guided free-fall	Buehler metallographic polishing table with a brass table covered with 150 grit Silicon Carbide paper rotating at either 730 or 1460 rpm prior to test. The table was allowed to free-wheel as the headform is dropped, so it could be decelerated in the impact.	Headform slice falls along guide rail to impact rotating table resulting in axial rotation of the headform slice	Angular acceleration. Impact force. Helmet deflection.	Figure S19
Pang (2011) [29] McIntosh (2013) [32] (2013b) [31]	Motorcycle	Hybrid III	Hybrid III neck attached to guided drop carriage	Guided free-fall	Horizontally moving aluminium striker plate covered with a high-friction outdoor surface tread.	Guided fall of headform onto horizontally moving impact surface	Linear and angular accelerations. Neck loads.	Figure S20
Dau (2012) [122] Hansen (2013) [37]	Bicycle	Rigid ISO/DIS 6220	Hybrid III neck attached to drop rail	Guided free-fall	30º angled steel anvil.	Guided drop of headform onto angled surface	Linear and angular accelerations. HIC. FE model criteria. Neck loads.	Figure S21
Giacomazzi (2009) [123]	Hockey	Hybrid III	Hybrid III neck attached to small trolley on linear bearing track	None	Impact pendulum with flat impact face weighing 70 kg.	Impact pendulum swung to impact headform	Linear and angular accelerations. GAMBIT HIP FE model criteria	Figure S22

Jadischke (2016) [124]	Football	Hybrid III	Hybrid III neck attached to linear slide table	None	Helmeted Hybrid III headform mounted to a 1.58 m pendulum arm via a rigid neck, total pendulum weight of 31 kg.	Impact pendulum swung to cause helmet to helmet impact of headforms	Linear acceleration. Change in linear velocity. Change in momentum. Neck loads.	Figure S23
Newman (2005) [113] Pellman (2006) [125]	Football	Hybrid III	Hybrid III neck attached to rigid platform or sliding table	None	Impact pendulum with weighted hammer. Impact surface has convex face covered with a layer of polycarbonate.	Impact pendulum swung to impact headform	Linear and angular accelerations. HIC Severity Index Neck loads.	Figure S24
Rowson (2015) [126]	Hockey	Medium NOCSAE	Hybrid III neck attached to linear slide table	None	Impact pendulum with rectangular arm 1.9 m long, total mass of 36.6 kg with 16.3 kg at the impact end. Flat, rigid, nylon impactor face with 12.7 cm diameter.	Pendulum swung to impact headform	Linear and angular accelerations	Figure S24
Tyson (2018) [127]	Football	Medium NOCSAE	Hybrid III neck attached to 5-degree-of-freedom Biokinetics slide table with a 16 kg sliding mass.	None	Pendulum impactor with 1.9 m long arm, total mass 37 kg with 15.5 kg impacting mass at the end. Nylon impactor face 20.3 cm in diameter with 12.7 cm radius of curvature.	Pendulum swung to impact headform	Linear and angular accelerations.	Figure S24
Bartsch (2012) [128]	Football	NOCSAE	Hybrid III neck attached to a rigid support	None	Swinging pendulum with NOCSAE headform fitted with a Riddell VSR-4 helmet, weighing a total of 6.2 kg.	One headform swung as pendulum to impact other, stationary headform	Linear and angular accelerations. Angular velocity. FE model criteria. Neck loads.	Figure S25
Pellman (2006) [125]	Football	Hybrid III	Hybrid III neck attached to torso which was mounted on a translating joint and table	None	Hemispherical surface with polycarbonate layer.	Impactor pneumatically driven to impact headform	Linear and angular accelerations. Neck loads. Chest accelerations.	Figure S26

Gwin (2010) [129]	Football	NOCSAE	Hybrid III neck attached to linear bearing table	None	Linear impactor head weighing 13.3 kg with a convex face padded with polyurethane foam.	Impact ram pneumatically accelerated to contact the headform	Linear acceleration. Severity Index.	Figure S27
Rousseau (2009) [130] (2009b) [131] (2009c) [132] Walsh (2012) [133]	Hockey	Hybrid III	Hybrid III neck attached to sliding table with 12.8 kg mass	None	Linear impactor with vinyl nitrile 602 foam disc and hemispherical nylon cap weighing a total of 16.6 kg.	Impactor pneumatically driven to impact headform	Linear and angular accelerations	Figure S27
Post (2013) [134] (2014) [135]	Hockey	Hybrid III	Hybrid III neck attached to sliding table with 12.8 kg mass	None	Linear impactor with MEP disc and hemispherical nylon cap weighing a total of 16.6 kg.	Impactor pneumatically driven to impact headform	Linear and angular accelerations. FE model criteria	Figure S27
Clark (2015) [116] (2016) [117]	Hockey	Hybrid III	Unbiased neckform on 12.8 kg sliding table	None	Nylon disc covering vinyl nitrile 602 foam. Nylon disc covered with vinyl nitrile R338 V foam and a Reebok 11 k shoulder pad.	Impact opponent pneumatically driven to impact headform	Linear and angular accelerations. Impact duration. FE model criteria.	Figure S27
Post (2012) [136]	Football	Hybrid III	Hybrid III neck attached to sliding table	None.	Linear impactor with vinyl nitrile 602 foam and nylon cap weighing a total of 13.1 kg.	Impactor pneumatically driven to impact headform	Linear and angular accelerations. FE model criteria	Figure S27
Hoshizaki (2016) [137]	Football	Hybrid III	Hybrid III neck attached to sliding table	None	Linear impactor head weighing 13.1 kg with hemispherical nylon cap and 1.5 in. MEP Shore 60A layer.	Impactor pneumatically driven to impact headform	Linear and angular accelerations	Figure S27
Pellman (2006) [125] Beckwith (2012) [138] Viano (2012b) [139] Jadischke (2016) [124]	Football.	Hybrid III	Hybrid III neck attached to sliding table	None	Linear impactor cap with vinyl nitrile foam covered by hard Nylon cap with convex shape of radius 127 mm. Total ram mass was 14 kg.	Impact ram pneumatically driven to impact the headform	Linear acceleration. Change in linear velocity. Change in momentum. Angular velocity. Neck loads.	Figure S28

Funk (2017) [140]	Football.	Hybrid III	Hybrid III neck attached to sliding carriage of mass 17.7 kg	None	Spherical extruded nylon 6/6 cap attached via Velcro to 1⅝" thick vinyl nitrile foam attached via Velcro to 14.3 kg linear ram	Impact ram accelerated to desired speed then disengaged from propulsion source so it strikes the specimen while coasting.	Linear acceleration. Linear velocity. Angular acceleration. Angular velocity. HIC. Combined metric.	Figure S27
Rowson (2008) [141]	Football	Hybrid III	Hybrid III neck attached to full Hybrid III dummy	None	Linear impactor with impacting characteristics of a typical football helmet	Impactor pneumatically driven to impact headform	Linear acceleration. Angular velocity. Impact force. Neck loads.	Figure S29
Viano (2012) [142]	Football. Mouthguards.	Modified Hybrid III with articulating mandible	Hybrid III neck attached to linear slide table	None	Linear impactor with foam and plastic cap assembly which simulates an impacting football helmet.	Impactor pneumatically driven to impact headform	Linear acceleration. Severity Index. HIC-36 Mandible forces.	Figure S30
Johnston (2015) [143]	Football	Rigid magnesium	Attached to a sliding table by a joint that allowed transverse plane rotation	None	Done-shaped nylon impactor weighing 14 kg.	Impactor pneumatically driven to impact side of facemask of helmeted headform	Linear and angular accelerations. Severity Index. Angular velocity. Angular displacement.	Figure S31
Kis (2013) [144]	Hockey	NOCSAE	Attached to specifically designed neck attachment that restricts rotational movement to a single axis with different neck attachments used for each of the three axes. The neck attachment was mounted to a rigid frame.	None	Impact piston, not further specified.	Impact piston pneumatically driven to impact headform	Linear and angular accelerations	Figure S32

McIntosh (2015) [145]	Boxing headguards and glove	Hybrid III	Hybrid III neck attached to a massive stand	None	Spring driven linear impactor guided by linear bearing weighing approx. 3.9 kg. Impact heads included a cylindrical mallet covered by a boxing glove and a disc-pad (not further specified).	Spring-loaded linear impactor driven to impact headform	Linear and angular accelerations. HIC-15 Impact force.	Figure S33
Ivarsson (2003) [146] King (2003) [147]	Football. Foam padding.	Hybrid III	Hybrid III neck attached to mini-sled	Propelled horizontally	Angled aluminium plate at 30º from vertical covered with various foam samples.	Mini-sled with upright headform and neck pneumatically accelerated to desired velocity and decelerated by the crushing of aluminium honeycomb with headform impacting impact surface	Linear and angular accelerations. Angular velocity. HIC36	Figure S34
Tyson (2018) [148]	Soccer	NOCSAE	Hybrid III neck attached to a 16 kg sliding mass	None	NOCSAE headform attached to Hybrid III neck attached to a 16 kg sliding mass	The striking headform, neck and sliding mass is propelled toward the struck headform using a cable and pulley system	Linear and angular accelerations.	Figure S35
Alem (1980) [149]	Industrial	Hybrid III	Hybrid III neck attached to linear rails	None	A 9 lb drop weight.	Drop weight released in free-fall or guided free-fall to hit the top of the headform causing a plunging motion of the head along the spinal (axial) direction	Linear acceleration. Impact force/acceleration. Neck loads.	Figure S36

Bartsch (2012) [150](2012b) [151]	Boxing headgear and glove. MMA glove.	Hybrid III	Hybrid III neck attached to Hybrid III dummy without lower extremities seated in a chair and secured with tie-down straps and 3500 N of sandbags	None	Steel sphere impact pendulum of 3.6 kg.	Impact pendulum swung to impact the headform	Linear and angular accelerations. Angular velocity. Neck loads.	Figure S37
O'Sullivan (2013) [152]	Taekwondo	Hybrid II	Hybrid II neck mounted in an inverted position to a free-hanging iron pendulum	None	Bowling ball weighing 6.75 kg suspended by a steel-lined cable.	Bowling ball swung to impact the headform	Linear acceleration	Figure S38
Withnall (2005) [112]	Soccer	Hybrid III	Hybrid III neck on 15.8 kg sliding table	None	FIFA inspected size 5 soccer ball, 430 g, 0.8 bar inflation pressure	Ball fired from air cannon as a projectile into headform	Linear acceleration. HIP.	Figure S39
Clark (2015) [116] (2016) [117]	Hockey	Hybrid III	Unbiased neckform on 12.8 kg sliding table	None	Hockey puck.	Puck fired from pneumatic launcher into headform	Linear and angular accelerations. Impact duration. FE model criteria	Figure S39
McIntosh (2003) [153]	Baseball. Hockey. Cricket.	Hybrid III 5^{th} percentile female	Hybrid III 5^{th} percentile female neck	None	Cooper CVB official league solid core centre baseball. Kookaburra Comet cricket ball. Vegum "official" hockey puck.	Impact projectiles fired using an air cannon into headform	Linear acceleration	Figure S39
Goldsmith (1982) [154]	Baseball	Valgem Corp headform	Coil spring simulating the neck which terminated in a flat plate attached to a 22.7 kg metal block	None	Little League baseball with mass of 148.2 g and average diameter of 7.23 cm.	Baseball fired from air cannon at helmeted headform which was suspended in an inverted position	Linear acceleration. Impact duration. Impact force. Contact area. Impact stress.	Figure S40

Viano (1993) [155]	Baseball	Hybrid III 5^{th} percentile female	Hybrid III 5^{th} percentile female neck attached to full Hybrid III 5^{th} percentile female dummy in a seated position	None	Commercially available baseballs.	Baseballs pneumatically fired to impact the headform	Linear acceleration. HIC. Neck loads.	Figure S41
Yang (2014) [156]	Baseball	Hybrid III	Hybrid III neck attached to full Hybrid III dummy seated, constrained in a chair fixed to the floor.	None	Baseballs meeting the weight and circumference specifications of the NOCSAE standard.	Baseballs fired using JUGS Curveball Pitching Machine to strike the forehead or left temple of the headform.	Linear and angular accelerations. HIC. SI.	Figure S41
Stepan (2010) [157]	Watersports	Hybrid III	Hybrid III neck attached to full Hybrid III dummy in seated posture	None	Metal boom from a J24 sailboat mounted horizontally onto a vertical rotating pole.	Boom accelerated to reproduce angular velocities seen during jibing to contact the occipital region of the headform, six feet away from the axis of rotation.	Linear acceleration.	Figure S42
Ghajari (2011) [158]	Motorcycle	Hybrid III	Hybrid III neck attached to full Hybrid III dummy in seated posture	Free-fall	Flat anvil.	Free-fall of dummy for frontal head impact into flat impact surface	Linear acceleration. Impact force. Neck loads.	Figure S43
Hering (2000) [159] Chinn (2001) [93] Ghajari (2011) [158]	Motorcycle	Hybrid III	Hybrid III neck attached to full Hybrid III pedestrian dummy	Free-fall	Flat anvil and oblique anvil at 15º from vertical covered with abrasive paper (grade 80 closed-coat aluminium oxide)	Suspended dummy released to free-fall with the first impact being the headform onto the anvil	Linear and angular accelerations. Impact force.	Figure S44
Aldman (1978) [160]	Motorcycle	Ogle-Opat	Ogle-Opat neck attached to full 50^{th} percentile male test dummy	Free-fall	New unpolished asphalt concrete surface Ab8t.	Dummy suspended from the side of a moving test cart released in different orientations to strike the impact surface	Linear and angular accelerations	Figure S45

Corner (1987) [161]	Motorcycle	GM Hybrid Head	GM Hybrid Neck attached to block of wood with wood enclosed in sand filled canvas	Propelled on test sled by a falling weight and pulleys	Various points on a car body. Laboratory floor.	Test sled accelerated to 12 m/s, then decelerated over 0.5 m, launching the surrogate at the impact opponent	Linear and angular accelerations	Figure S46
Hodgson (1990) [162]	Bicycle	Hybrid III	Hybrid III neck attached to full Hybrid III dummy torso	Propelled horizontally	Concrete barrier inclined at angles of 30º, 45º, 60º and 90º.	Dummy driven horizontally in-line with body longitudinal axis by spring propulsion so head impacted angled surface with dummy neck horizontal	Linear and angular accelerations. Impact force. Neck loads.	Figure S47
Scher (2006) [163] Richards (2009) [164]	Ski or snowsport	Hybrid III	Hybrid III neck attached to full Hybrid III dummy in standing posture on snowboard	Guided fall along cable path followed by backward fall of standing dummy	Snow.	Dummy released to fall along declining cable path while upright, exiting the cable at a mound of snow to cause the "opposite-edge" snowboarder trip mechanism causing the dummy to fall backward	Linear acceleration. HIC. Neck loads.	Figure S48
Scher (2008) [165]	Ski or snowsport	Hybrid III 10-year-old	Hybrid III 10-year-old neck attached to full Hybrid III 10-year-old dummy in standing posture	Swinging pendulum	Bar stock mounted for chest impact (inertial head loading). Wooden pole. 5th percentile female Hybrid III full dummy.	Dummy swung like a pendulum in the AP axis direction to impact against impact opponent while in the standing position	Linear acceleration. HIC. Neck loads.	Figure S49
Yamaguchi (2014) [166]	Climbing	Size J rigid magnesium. Hybrid III.	Hybrid III neck attached to full Hybrid III dummy with standing pelvis	Swinging pendulum	Vertical steel barrier.	Dummy swung in the AP axis and transverse axis directions to impact opponent while horizontal to the ground	Linear acceleration. HIC. Neck loads.	Figure S50

Yamaguchi (2014) [166]	Climbing	Size J rigid magnesium. Hybrid III.	Hybrid III neck attached to full Hybrid III dummy with standing pelvis	Upright free-fall	Steel slotted test bed.	Dummy released to free-fall while upright to impact surface first with feet, followed by the buttocks, shoulders and then occipital head	Linear acceleration. HIC. Neck loads.	Figure S51
Newman (2005) [113] Pellman (2006) [125]	Football	Hybrid III	Hybrid III neck attached to full dummy	Guided suspended free-fall using a gantry	Full Hybird III dummy in guided, suspended free-fall from a gantry.	Two gantries suspended two Hybrid III dummies which were released in guided free-fall to impact each other at a predetermined impact site and velocity	Linear and angular accelerations	Figure S52

[c] Unless otherwise specified, reference to anthropometric test devices, such as the Hybrid III, refer to the 50^{th} percentile adult male.

[d] HIC – Head Injury Criterion, RIC – Rotational Injury Criterion, PRHIC – Power Rotational Head Injury Criterion, GAMBIT – Generalized Acceleration Model for Brain Injury Threshold, HIP – Head Impact Power, FE – finite element, BrIC – Brain Injury Criterion.

3.1 Impact condition

There were three ways of producing a head impact with an impact opponent in the new headgear test methods. The first used pre-impact motion of the headform to contact a stationary impact opponent. The headform is either dropped vertically under gravity or propelled using a spring, pendulum or mini-sled. The second type of impact involved a moving impact opponent contacting a stationary headform. Impact opponents were propelled via cannons, pneumatic linear impactors, pendulums or fell due to gravity to impact an initially stationary helmeted headform. The final type of test performed an impact while both the headform and the impact opponent were in motion, for example the headform falling onto a translating or rotating surface, or two test surrogates propelled into contact with one another.

There was a wide range of impact opponents in the reviewed new headgear tests including roadway surrogate surfaces, snow, other helmeted headforms, various impactor tips, projectiles as well as rigid anvils and MEP surfaces referenced in current certification standards.

The impact orientation was variable in most new tests and involved two components: 1) the orientation of the headform with reference to the impact opponent and, 2) the relative velocity vector between the test surrogate and the impact opponent. Common strategies for achieving different impact orientations included changing the pre-impact headform position relative to the impact opponent, varying the angle of the impact surface and changing the point of impact on the helmet.

3.2 Human head test surrogate

Headform types used in the reviewed new test methods ranged in complexity from a metal 2D cylinder to rigid metal humanoid headforms, manikins, the NOCSAE headform and headforms from automotive ATDs including the Ogle OPAT, the Hybrid II and the Hybrid III of various sizes.

The new headgear test methods applied various types of boundary conditions to the test headform at impact. Many studies affixed the headform to a flexible neck while others applied no boundary conditions to the head at impact. Studies using a neck variously attached the lower neck to a drop carriage, drop rail, translating table, cantilevered mass, ATD torso or full ATD. The types of necks and ATDs were predominantly from automotive applications and included the Ogle OPAT, motorcycle ATD and Hybrid III of various sizes. Other non-rigid neck surrogates included the so-called “unbiased neckform”, simple uniaxial mechanical joints and a coil spring.

3.3 Assessment criteria and thresholds

All of the new testing studies used linear acceleration of the headform as an assessment criterion, see Table 2. Additional kinematic assessment criteria included angular acceleration and angular velocity of the headform, the Severity Index, the Head Injury Criterion (HIC), the Rotational Injury Criterion (RIC) [167], the Power Rotational Head Injury Criterion (PRHIC) [167,168], the Generalized Acceleration Model for Brain Injury Threshold (GAMBIT) [169], and Head Impact Power (HIP) [170]. Further criteria related to headform motion utilised finite element models of the head. These criteria involved applying a measured headform response to a rigid skull and deformable brain computational model. Headgear assessment compared finite element outputs such as stress, strain and strain rate developed within the simulated brain tissue [97,134,136,171–173]. Load criteria used to assess headgear in the new impact tests included impact forces, mandible forces and neck loads. Other headgear assessment criteria included impact duration, contact area, impact stress and helmet deflection.

In the new test methods, pass/fail thresholds for allowable headform response were rarely specified. Typically, these studies provided some estimate of head injury risk in the test impacts by referring to previously published studies, such as human cadaver impact experiments [174], animal studies [25,175], and tissue level models [176–178]. Established head injury risk curves were also referenced, such as those developed in automotive applications for specific surrogates such as the Hybrid III [114,120,162,163,166,179], or developed through reconstruction of real-world helmeted football incidents [172,180–183].

4. Discussion

4.1 Requirements of a standard

The new headgear test methods reviewed in Table 2 are primarily research studies incorporating greater complexity in the impact condition and exploring more potential headgear assessment metrics than impact tests in the current standards summarised in Table 1. Most new impact test methods are unsuitable for a standards test in their current form due to their complexity. An impact test suitable for certification standards should be simple, robust and capable of producing repeatable and reproducible results [184]. For a given test method, the same results should be achievable for commercial test houses, headgear manufacturers and academic researchers in order to facilitate the processes of ongoing headgear development and certification.

Practically, impact tests in standards should be only as complicated as is needed to adequately represent the injurious event. Often the process of defining what is adequate can take considerable research. For example, a series of studies published by Pellman and colleagues [109,110,125,185,186] analysed and reconstructed mTBI-producing impacts that occurred in football games using full reconstructions involving two complete ATDs. These tests and others were used as a precursor to developing a simpler laboratory impact test method that simulates helmet-to-helmet impacts (Fig. S28) [125,138]. Since the resulting pneumatic linear impactor test has acceptable repeatability, reproducibility and practicality, it is being added to the NOCSAE standard for football helmets, effective November 2019 [187,188]. This test development process highlights a difference between the aims of a headgear researcher, who may want to recreate the injury event as realistically as possible, compared to the needs of a standard test, where some realism is sacrificed to achieve other benefits.

Sensitivity is another important aspect of a standards test. If the impact condition, test surrogate and assessment criteria are not sensitive to the injury related conditions, essentially being unable to distinguish between a protective and non-protective (or less-protective) headgear, the test does not achieve what is intended.

The goal for new impact tests in headgear standards is to create the simplest, most repeatable set of tests that capture the greatest number of injury related risk factors and thus results in the greatest reduction of injuries. The ideal situation would use high fidelity impact and injury research data to contrive a simple standards test that was ultimately entirely effective at stratifying tested headgear as to their relative ability to protect against defined severities of mTBI and severe TBI/skull fracture. The following sections discuss how new headgear research methods and other studies can inform development of standards-appropriate impact tests.

4.2 Impact condition

There is significant variation in the types of head impacts that occur in different sports and within the same sport yet impact tests in certification standards use very similar impact test dynamics across sports (Table 1). New laboratory impact tests have greater impact condition variation and from a research perspective, accurate definition of real injurious impact conditions is critical for recreating realistic head impact scenarios in the lab. Of importance are the pre-impact orientation of the head relative to the impact opponent, the relative velocity vector between the head and the impact opponent, and the mechanical response of the impact opponent.

Research efforts to define the impact condition in sports head impacts have utilised a number of different techniques and information sources. In football, in-game head impact events were analysed and reconstructed through video analysis [109,113,189]. For bicycle and motorcycle helmets, headform impact orientations and the relative velocity vector between the headform and the impact opponent have been informed by crash investigations [93,97,190]**.** Impact orientation and severity have also been estimated based on the residual damage of single impact helmet liners such as expanded polystyrene liners commonly used in bicycle and motorcycle helmets [191,192]. However, using residual deformation to estimate impact severity is challenging, has high variability and requires data on specific helmet models making it potentially unreliable for widespread application [192].

Definition of the head impact condition in other sports suffers from a paucity of available real world head impact data. Multibody simulations and parametric studies have been used to fill this void by estimating unknown parameters that describe the impact condition such as head impact location and velocity [193]. A technique called model-based image matching has been used to approximate head impact velocities in an alpine skiing crash from uncalibrated video sequences [194]. Others have directly measured head impact events during sports using instrumentation mounted on the helmet, skin or within mouthguards, reducing the reliance on physical or simulated incident reconstruction [195–204]. These efforts inform researchers about the head impact orientation and resulting kinematics in specific sports but are not without limitations. Imperfect coupling between the impact measurement device and the skull introduces inaccuracies in some systems [205–208] and false positives can be common [209,210], meaning video confirmation of head impact events is important [211]. High quality, detailed field and clinical data is essential to ensuring headgear impact tests are relevant to injurious scenarios that occur in the real world.

For a standards test, it is not necessary to recreate all injurious head impact scenarios if the headgear response in one injurious impact type correlates to the headgear response in another. Exploratory testing can define the minimum number of independently responding injurious impact configurations, reducing the number of required test configurations. Test method complexity can also be reduced through exploratory testing. For example, the response of a falling test headform impacting a translating roadway surface surrogate can, in some circumstances, be replicated with the easier to manage test setup of a falling headform impacting a stationary oblique-angled anvil [212].

Details regarding the impact opponent are typically easier to define than the impact dynamics but the number of potential impact opponents in different sports can be vast. Controlled in-field testing of impact opponents is needed to capture the range of conditions for variable impact surface conditions such as snow or loose-fill materials. Development of standards-suitable repeatable impact surface materials can follow. Previous examples of this include the pneumatic linear impactor face which was designed to simulate helmet-to-helmet contacts in football [125] and the MEP pad which provides a reproducible impact surface surrogate to simulate impacts with the ground [75,213,214].

In standards testing, rigid steel impact surfaces are common and represent worst-case impact opponents for linear impact evaluation due to the maximized potential for fully compressing headgear padding. Rigid surfaces, however, may not represent the worst-case scenario when considering angular headform response and consideration of other surfaces will be necessary for new standards tests. King et al. [215] performed head impact tests using a Hybrid III head and neck on a mini-sled and found that, in four out of nine front impact tests with a bicycle helmet and two out of nine tests with a football helmet, added compliance of the helmet increased the rotational acceleration of the headform compared to the unhelmeted condition. In another study, hockey shoulder pads added to a pneumatic linear impactor tip at an impact velocity of 6.5 m/s reduced the peak linear acceleration experienced by the headform (151.9 *g* to 100.9 *g*) but increased the peak angular acceleration (7.4 krad/s^2 to 8.2 krad/s^2) compared to a bare impactor condition [216]. Exploratory testing is needed to ensure higher injury risk impact opponents are identified and accounted for in a standards test.

The high incidence of sports-related mTBI despite headgear protection has generated discussion regarding the impact severity at which sports headgear are tested. Most standards call for headgear to be tested in impacts at or near the highest levels of severity deemed reasonable [217]. Headgear that meet the required response in these impacts are considered protective for all impacts of equal or lesser severity, although they are tested at only the one high impact severity [217]. This approach might ignore headgear performance in lower severity impacts and mTBI is often sustained in impacts that exhibit resultant peak linear head accelerations well below the threshold level allowed in certification standards [195,218–220]. Bicycle, motorcycle, football and martial arts headgear subjected to linear impact tests at different severities show an approximately linear relationship between impact severity and peak linear acceleration of the headform up to bottoming-out of the protective liner [150,217,221–223]. Such a relationship suggests that indeed the linear response of these headgear at lower levels of impact is largely determined by the single, higher impact severity threshold [217]. Whether this linear relationship at increasing impact severity holds for angular headform response in impact tests with additional headform degrees of freedom and across various head impact configurations, impact opponents and for emerging helmet technologies that leverage different energy attenuating strategies to crushable foam is unknown.

A relatively recent issue for sports headgear testing is the increasing use of aftermarket helmet accessories, such as cameras or add-on caps. Cameras influence interaction between the helmeted head and the impact opponent, since camera-mounting points alone project further from the headgear surface than is allowed in many headgear standards. It is also unclear whether the camera-mounting points are designed to collapse or break on impact [114]. A laboratory study investigating this issue by performing flat surface impacts to a surrogate camera mounted on a bicycle helmet found that the presence of a camera altered the kinematics and forces experienced by the headform. The average risk of severe concussion, using Cumulative Strain Damage Measure-25 injury risk curves [172], was reduced in 4 m/s impacts with the camera attached (from 25% to 7%), but was increased, on average, in 6 m/s impacts (from 18% to 58%) [114]. The effect these accessories have during oblique impacts where they can potentially snag on the impact opponent has not been methodically assessed. Aftermarket helmet add-on caps have been developed with the aim of mitigating concussions in football but change the impact response of the helmet. According to NOCSAE, these additions create a new and untested helmet model, as defined in NOCSAE standards, and therefore make the certification of previously certified helmet models voidable [224]. Published impact test results for add-on caps are scarce but, even within the limited test data available, additional padding applied to the helmet exterior may not always reduce the severity of the impact in a drop test [225,226]. Developing separate standards for different types of headgear accessories would help better understand their effect on injury risk and control their influence in sport head impacts.

4.3 Human head test surrogate

Rigid headforms are not designed to respond like a human head to impact but most often have a defined shape based on human anthropometry. The NOCSAE urethane headform and Hybrid III headform have a humanlike shape, mass, moment of inertia and are designed to respond like the human head in certain impact conditions [227–229]. However, the NOCSAE and Hybrid III headforms are not interchangeable. The NOCSAE headform is considered more anatomically accurate than the Hybrid III, particularly at the base of the skull, cheeks, jaw and chin [230,231]. The anatomical inaccuracies of the Hybrid III headform are a limitation for headgear assessment since the chin and nape, in particular, affect the fit and retention of headgear and may therefore affect the helmeted impact response. The NOCSAE headform was originally designed to be mounted on a rigid arm whereas the Hybrid III headform is compatible with the Hybrid III neck, although a method now exists to affix the NOCSAE headform to the Hybrid III neck [125,230]. The Hybrid III is the most extensively used humanoid headform in biomechanical research and the earliest to incorporate a system of measuring both the linear and angular kinematics in an impact [230]. Head injury risk estimates have been developed with the Hybrid III headform which might be a reason attracting headgear researchers to use the Hybrid III rather than other surrogates [174].

It is appealing to evaluate headgear using a surrogate designed to respond like the human head to impact. Hence, the biofidelic impact response of the Hybrid III is commonly referenced as justification for it being used in the reviewed new headgear tests [34,93,112,179]. However, the biofidelity of the Hybrid III headform for headgear impact testing has relatively little validation. The Hybrid III headform was originally developed to respond like the human head in unprotected, short duration impacts against rigid anvils rather than in helmeted impacts [228]. In a study that compared the linear impact response of the Hybrid III and rigid headforms to the response of human cadavers in motorcycle helmeted impacts, the rigid DOT headform, rather than the Hybrid III headform, most closely resembled the cadaver response [191]. The NOCSAE headform was designed to have similar static load/deflection characteristics of the human cadaver skull and humanlike impact response in a small series of helmeted and athletic turf impacts [227]. Despite both being based on human cadaver head response, the NOCSAE and Hybrid III headforms provide significant differences in peak linear and angular accelerations in comparative impact tests [232,233]. Given these differences, the specific test methods and impact locations of any new or broader standard for sports headgear will determine which of these headforms is more suitable. Indeed, it is also possible that neither headform is valid for the task and that a new headform will be needed to properly assess how headgear attenuates linear and angular kinematics and their respective injury risks.

Another issue related to headforms for headgear evaluation is the available sizes. While rigid headforms have the most extensive array of sizes at circumference increments of 10 mm [84,85], size variations of the Hybrid III and NOCSAE are limited. At least three more Hybrid III sizes are needed to cope with all helmet sizes [234], and NOCSAE standards provide exceptions or specify procedures to deal with poor fit as described in Section 2.2 of this review. The issue of humanlike impact response is further complicated when considering paediatric size variations. The head impact response of current paediatric Hybrid III ATDs does not agree with age-matched unprotected human skull cadaver impacts [235]. Corresponding human paediatric head impact response data is sparse, particularly for padded impacts, but is needed to create biofidelic paediatric head surrogates for headgear impact testing.

With increasing emphasis on the angular response of the protected headform in an impact, a critical concern is the surface characteristics of the headform, or more generally, the friction at the interface between the helmet and the headform. For instance, covering a rigid magnesium headform with 1 mm thick silicone rubber increased peak rotational acceleration from 6.1 rad/s^2 to 11.6 rad/s^2 (89% increase) compared to the uncovered headform in a free-flight headform drop onto a 30º oblique anvil [99]. None of the headforms currently used for headgear testing have surface friction representative of dry or sweaty human hair or skin. A recent study identified that headforms do not include scalp-skull friction and therefore there is no tensioning effect of the skin [236]. Furthermore, the coefficient of friction at the human cadaver scalp and helmet liner interface (0.29) was significantly different to the Hybrid III headform and helmet liner interface (0.75), and to the rigid magnesium EN960 headform and helmet liner interface (0.16) [236]. In past attempts to address this issue, researchers have made surface modifications to test headforms by addition of an artificial scalp and wig [28,92], a layer of PVC plastisol [35], silicon rubber [99], or two layers of nylon stocking material [113,207]. The fidelity of these modifications to human skin and hair has not been demonstrated.

Headform inertial properties are a further important contributor to the angular response in helmeted impacts. The moments of inertia of automotive ATD headforms are within the wide range reported for human heads although considerable differences exist between headforms (see Table 3) [237–239]. In the typical head reference frame, the human head products of inertia are non-zero, since this reference frame is not aligned with the principal axes of the head, but are not reported in literature. It is not known whether automotive ATD headform products of inertia match the human head and the degree to which incorrect properties will influence the headform response in a protected sports head impact.

Table 3 Mass and inertial properties comparison between human cadaver data and automotive ATD headforms.

Head/ headform [ref]	Mass (kg)	Ixx (kg-cm^2)	Iyy (kg-cm^2)	Izz (kg-cm^2)	Ixy (kg-cm^2)	Ixz (kg-cm^2)	Iyz (kg-cm^2)
Human cadaver [237,238]	4.5	211	231	179	-	-	-
Hybrid III [239]	4.54	153	210	181	180	198	198
NOCSAE [239]	4.51	183	240	167	207	192	200
ES-2re [238]	4.00	147	193	163	-	-	-
WorldSID [238]	4.24	189	159	149	-	-	-

The choice of human head surrogate for headgear impact testing in standards could look beyond currently available test devices. A simple head surrogate may be appropriate if other factors such as the assessment criteria, headform boundary conditions and impact dynamics are well defined. For instance, if mTBI relates to angular velocity change of the head in an impact, an ellipsoid that mimics the inertial properties of the head and neck as well as the surface friction and geometry of the head/helmet interface could provide adequate headgear assessment of relative mTBI risk.

4.4 Head surrogate boundary conditions

One of the criticisms of headgear impact testing in standards is the lack of rotation since the guided drop test restricts motion of the headform in all but one axis (Fig. 1). Every research impact test summarised in Table 2 allowed additional headform degrees of freedom, though there were considerable differences among the boundary conditions applied to the head.

One major question is how the neck influences the kinematics of the headform in a protected head impact and whether headgear impact tests need to include a neck. A number of studies suggest the neck plays only a small role in helmeted head impact response. In a finite element simulation study of 24 bicycle helmet impact configurations and three helmet conditions (no helmet, road bicycle helmet, skate bicycle helmet) analysed for durations up to 15 ms after impact, the detached head, on average, produced 6% higher peak linear acceleration, 8% higher peak angular acceleration, 5% higher peak angular velocity and 4% higher peak brain tissue strain compared to the head attached to a neck and body [240]. In a simulation study reconstructing bicycle crashes using multibody analysis, the head was regarded as mechanically separated from the human body for the first 2-3 ms of the impact such that the neck has a negligible influence of the head response in this period of time [241]. Furthermore, Willinger et al. [190] showed experimentally that the angular acceleration response of the detached Hybrid III head protected by a bicycle helmet is similar to the response attached to the Hybrid III neck for the first 10 ms of the impact, admitting the neck influences head kinematics for longer durations after the impact. Axial and oblique padded impacts to the head of cadaver head and neck preparations have shown a time delay of up to 9 ms between force generation at the point of impact (head to padded anvil) and forces measured at the lower neck (T1), possibly suggesting mechanical separation and minimal influence of the neck on head kinematics for this time duration [242]. Given the above studies, and the fact that human cadaveric specimens exhibit an atlanto-occipital neutral zone (joint motion with no force) in the range of 10 degrees, Willinger et al. [234] reason that headgear impact testing without a neck is valid for certain short-duration (5-10 ms) impact configurations resulting in up to 10 degrees of headform rotation. European Working Group 3 within COST TU1101 is developing a bicycle helmet impact test method without a neck using these assumptions [234].

On the other hand, several studies suggest that the neck has a significant influence on head kinematics, particularly the rotational response, during protected head impacts. Greater angular accelerations of the head were predicted for simulated helmeted jockey incidents using only the head compared to the full body in a multibody modelling approach [243]. It was also noted that the direction of head acceleration can be altered by the absence of a neck [243]. Similarly, Beusenberg et al. [244] simulated four football helmeted impact configurations and varied the headform boundary conditions finding that neck coupling, while having a limited effect on the linear head accelerations, can reverse the direction of angular acceleration in some rotational axes. Physical impact testing of motorcycle helmets onto a flat anvil found that the influence of the neck and body is strongly dependent on the impact configuration [159]. In drops onto the parietal region of the head with the body oriented perpendicular to the flat impact surface, peak rotational acceleration was much greater using a complete dummy (mean of 5.3 krad/s^2 at 6.0 m/s) compared to the detached head (mean of 3.4 krad/s^2 at 6.0 m/s) thought to be due to the body dynamics transmitting large forces to the head through the neck [159]. Against an oblique anvil, angled 15 degrees from the direction of headform motion, peak rotational acceleration was also greater in full dummy drops compared to the detached head. Equivalent peak head kinematics to full body dummy head impacts at 6.0 m/s were achieved for detached headform impacts at between 6.0 and 7.5 m/s. The increased peak angular acceleration was due to the momentum of the body causing rotational motion about an axis in the neck area, motion that was not present in detached headform impacts. Contrastingly, peak rotational acceleration was lower using a complete dummy compared to the detached head for frontal head impacts with the body angled 30 degrees from the flat impact surface (mean values of 3.7 krad/s^2 vs 4.8 krad/s^2 at 6.0 m/s) and for rear head impacts with the body parallel to the flat impact surface (mean values of 4.3 krad/s^2 vs 5.6 krad/s^2 at 5.2 m/s). The freedom of movement of the impacting headform is reduced when connected to the body in these orientations, lowering the peak rotational accelerations [159].

The choice of whether a neck is necessary to accurately replicate helmeted head impact kinematics is therefore dependent on the impact condition and the assessment criteria of a proposed impact test. The impact condition in sports head impacts can depart from where the neck appears to have negligible influence on the headform response, for example at longer durations of 8-20 ms observed for head impacts against racetrack turf [243], or the average 15 ms duration of football helmeted head-to-head or head-to-body impacts resulting in concussion [109]. If rotational assessment criteria are needed in a longer duration impact test, perhaps against a less than rigid impact opponent, the influence of the neck on the headform kinematics cannot be ignored.

A further influence of the neck that is important for headgear testing is the effect on foam crush. Ghajari et al. simulated flat anvil drop tests of a motorcycle helmeted Hybrid III headform with and without the neck and body attached [158]. At 6 m/s the detached head experienced higher peak linear acceleration (133 *g* compared to 113 *g)* and lower liner crush (64% compared to 79%) than the full dummy attached headform. However at 7.5 m/s, greater peak linear head acceleration was experienced by the full dummy attached headform rather than the detached headform (278 *g* compared to 216 *g*) due to densification of the protective liner (91% crush compared to 81% crush). Greater liner crush due to the presence of the neck and body has also been demonstrated using human body models in impacts to an oblique surface and in impacts onto a flat surface with an initial oblique velocity vector [94,245]. Modifying the inertial properties of the detached headform for use in free-flight headform tests has been suggested to account for these differences in liner crush [94,158].

When headgear impact testing needs a neck, the accuracy of the recreated head motion is dependent on the biofidelity of the neck surrogate. The most common neck surrogate in the new impact test methods belongs to the Hybrid III ATD, a neck originally designed so that the flexion-extension motion of the Hybrid III headform matched that of volunteer and cadaver automotive sled test data [246]. Outside frontal and rear-end car crash applications, the Hybrid III neck has substantial biofidelity limitations. It is being increasingly used in headgear evaluation studies despite being too stiff and providing excessive resistance to horizontal translational motion between the head and torso [247,248]. The Hybrid III neck response corridors are not adequate to properly reproduce human motions of the neck, and therefore the head, in situations where load comes in multiple directions, such as rollovers [249] and so potentially the same is true for head impacts in sports. One finite element simulation investigation has suggested that, for certain helmeted impacts, the headform kinematics when attached to the Hybrid III neck could be less humanlike than without a neck altogether [234]. In this investigation, three helmeted impact configurations were simulated with a Hybrid III headform model and three boundary conditions: no neck, attached to a Hybrid III neck model, attached to a human cervical spine neck model by merging the skull base to the rigidly modelled aluminum Hybrid III headform base [234]. There are substantial caveats to the interpretation, namely that neither the Hybrid III nor human neck models had been validated to the simulated impact conditions and substantial divergence of the headform angular velocity was noted between the no neck and attached neck simulations at durations longer than 15 ms in two of three configurations [234].

Alternative mechanical neck surrogates have been used for headgear evaluation. Head impacts with a predominantly axial orientation have been replicated using a novel mechanical neck surrogate validated to the range of motion and stiffness data of human cadavers in flexion-extension rotation and axial compression [105,106,250,251]. A so-called unbiased neck is used in other studies, but the biofidelity of this surrogate has not been reported [252,253]. The motorcycle ATD neck has been used to test soccer headgear [112]. Without further validation, it is difficult to be confident these or other automotive neck surrogates are appropriate for headgear testing. This is also true of paediatric automotive neck surrogates. Thus there is an urgent need for biofidelic neck surrogates capable of accommodating multi-directional head impact events such as those that occur in sports.

Attachment of the lower neck is considered in a number of the new headgear impact test methods. Translating tables and cantilevered masses are used to simulate the mass of the torso in some studies [39,254], although the difference in headform response necessitating these features has not been quantified. A study simulating 4 configurations of a helmet to helmet impact in football found very little effect (not quantified) of changing the attached body mass from 5 kg to 50 kg on the resulting head kinematics [244]. In 4.4 m/s impacts imparted to the Hybrid III head and neck by a 3.6 kg pendulum, no significant difference was observed in linear or angular headform response for a rigidly mounted lower neck compared to the lower neck mounted to a 12.78 kg translating table [254]. Aldman et al. [120,160] compared drops of a complete ATD from a moving carriage to rail-guided dummy headform and neck drops onto a rotating disc, simulating the relative vertical and horizontal velocities between the dropped ATD and the roadway. Comparison of similar test configurations (occipital impact with inclined neck/body, unpolished/unworn road surface, polycarbonate shell helmet, horizontal velocity component of 8.1 – 8.4 m/s and vertical velocity component of 5.1 – 5.2 m/s) show peak linear headform acceleration of 120 – 150 g and peak angular acceleration of 5500 – 11400 rad/s^2 in the full dummy drops compared to 90 – 135 g and 8000 – 13500 rad/s^2 for the guided headform and neck drops onto the rotating disc [120,160].

Practically, a standards test for headgear is unlikely to use a complete ATD, which could potentially have the most humanlike boundary conditions for the head. Inclusion of a neck does not appear to hinder test practicality, as evidenced by the many ways a neck has been incorporated into new impact tests (Table 2), however there is a pressing need for appropriately biofidelic neck surrogates for headgear testing. Defining an appropriate headgear assessment criteria and consideration of the impact condition, particularly the impact duration, will determine for each specific sport what boundary conditions are needed for headgear impact tests in standards.

4.5 Assessment criteria and thresholds

The linear acceleration and time duration based criteria and thresholds predominantly used in current headgear certification standards (Table 1) appear to be based on human cadaver head impact experiments performed in the 1960s and formed the Wayne State University Concussion Tolerance Curve [3,255,256]. These experiments tested unprotected heads and observed the occurrence (or not) of skull fracture in short duration impacts, inferring the tolerance of concussion based on the clinical observation that concussion was almost always sustained in skull fracture cases [257]. Since then, biomechanical research has shown that the angular response of the head is also a determinant of the risk of TBI [20,24,25], leading to many other criteria being proposed for brain injury risk assessment. Headgear evaluation using rotational measures in the past has primarily focussed on angular acceleration, however current head injury evaluation in motor vehicle occupant protection is incorporating measurements of angular velocity [172,258]. At present, there is no consensus on which kinematic measure provides the best brain injury prediction, or prediction of a specific brain injury, for headgear impacts. Thresholds may also differ between youth and adults and vary across the paediatric age spectrum [259].

Acceleration and velocity are measures of the global kinematics of the head and may not capture the complex and local deformation patterns in the tissues of the brain during an impact. TBI is intimately related to brain tissue deformation and therefore assessment of the local response of the brain will likely provide better TBI risk assessment than the global head response [215]. Metrics based on the brain tissue response, such as strain or strain distribution, have been shown to better correlate with risk of TBI in reconstructed road traffic, football incidents and animal tests than traditional head acceleration measures [172,180–182]. Detailed finite element models of traumatic brain injury have allowed researchers to predict these metrics [260,261], and to show correlation between the regions with large tissue strain during impacts and the sites of injury after mild TBI [262].

The inclusion of an additional brain injury related criterion into future impact test standards appears inevitable. European Working Group 3 within COST TU1101 are considering either a global kinematic assessment variable, such as Brain Injury Criteria (BrIC) [172,182], HIP [170], RIC [167] or PHRIC [167,168], or finite element head model based injury risk assessment for use in bicycle helmet testing [234]. Football helmet evaluation in the NOCSAE standard will include a 6000 rad/s^2 threshold of allowable angular acceleration in linear impactor testing from November 2019 [188]. The introduction of new criteria is superfluous unless they provide improved headgear evaluation. In some head impact configurations, linear acceleration is closely correlated to the headform rotational response so adding an angular criterion is unnecessary [93]. In certain impact tests, finite element model outputs can be predicted through statistical models and the global kinematics of the headform meaning simulation in these configurations is unjustified [115,263].

Developing a threshold value for a headgear assessment metric is complex. The injury threshold that applies to humans is usually not the same as an injury assessment reference value (IARV) that applies to a human head surrogate. Both the risk of injury to a human and the response of the human surrogate in a defined impact condition are needed to develop IARVs. Previously generated IARVs, such as those for automotive occupants in crashes, relate to unprotected heads and their appropriateness for headgear evaluation has not been demonstrated. Collections of field data from helmeted football and hockey participants have been used to determine head injury risk specific to sports headgear impacts for headgear consumer information rating schemes [126,264]. New reliable in-field biomechanical head impact measurements are needed for continued injury risk refinement in these sports and for injury risk development in other sports. Well-validated instrumented mouthguards are currently available for this purpose [265–267]. Furthermore, injury risks or tolerances generated for use with one human head surrogate, such as a specific finite element head model, cannot be used with a different surrogate. Due to biofidelity limitations of available head and neck surrogates and the practical requirements of a standards test, the ability to determine the absolute risk of head injury in sports impacts may be both unachievable and unnecessary for headgear certification by standards. Determining the relative risk of injury through an appropriate metric can differentiate a more protective headgear from others. Using this approach, an effective threshold level can be initially set and refined using continued injury surveillance and field data.

The current lack of diagnostic precision for sports-related mTBI is an impediment to establishing a headgear assessment metric and threshold for this injury. Development of a biomechanical injury risk relationship requires impact response data and corresponding clinical data to confirm the presence or absence of injury. MTBI diagnosis presently relies on self-reporting and a subjective assessment of symptoms making it prone to variability in distinguishing injured from uninjured sports participants [268]. Moreover, TBI severity occurs along a continuum meaning not all mTBI injuries are equal. There is currently no objective measure to confirm mTBI incidence or stratify into TBI severities although research groups and private companies are exploring blood/fluid biomarkers, imaging techniques and vision based tests [269].

4.6 Basis for developing headgear impact tests for standards

Historically, headgear impact test development for standards prioritized and succeeded in preventing incidents of catastrophic head injury, such as skull fracture and severe brain injury, initially in motorized sports, bicycling, football and hockey. Further sport-specific standards developed by incorporating anvils that reflected hazards specific to each sport while the impact test methods are substantially the same, seen in Table 1. Exceptions are impact tests in more recently developed standards, such as ASTM F2439 for soccer headgear that was motivated by a high incidence of concussion in soccer and originally approved in 2006. The choice of test apparatus and failure threshold in F2439, different to the majority of other contemporary sports headgear standards, reflect the distinction in injury focus and the process that was undertaken defining the primary injury mechanism and injury condition [270].

The field of head impact injury and prevention now benefits from improved understanding of the brain injury mechanisms and the ability to define the response of the head in an impact quantitatively, whether through video analysis, in-field measurement, reconstruction or simulation. The path forward will utilize this progress to develop new impact tests. This review of current sports headgear standards and new headgear impact test methods highlights the important aspects of an impact test method for standards, which forms a set of steps for developing future standards summarised in Table 4. Table 4 also provides reference areas of research or impact testing strategies to support each step of the process.

A standards impact test will necessarily require compromise between exact replication of all possible situations that cause injury and the need for consistency. Analysis of in-field head impacts provides priorities for impact testing in standards through identification of the most common situations that result in injury and the situations that result in the most consequential injury. The most important features of in-field injurious head impacts to mimic in headgear impact testing are those that influence the risk of head injury. Identifying these features requires careful and thorough consideration of each step listed in Table 4.

A dilemma for setting an impact test is if we cannot replicate all hazardous conditions, we might unknowingly design a standard that can cause harm, but without developing or updating impact tests, we may be limiting improvements in head injury risk reduction for sports participants. If impact test development follows a thorough process, considering each important aspect listed in Table 4 and utilizing the corresponding research area and testing strategy, the risk of creating a standard that could unknowingly cause harm is minimised. In particular, comprehensive exploratory impact testing and ongoing injury surveillance of sports participants ensures a thorough understanding of how headgear performs in the field and the effect on injury outcomes.

At present and as always, there are limitations to what we know and our understanding of aspects of head impact biomechanics and prevention is continuously evolving. Table 4 provides a roadmap for researchers and those setting standards to identify what information is missing for headgear in specific sports and to work toward filling the gaps in knowledge.

Table 4 Sports headgear impact test development roadmap for certification standards.

Steps and considerations in impact test development	**Reference research area or strategy**
1) Identify prevalent or significant head injuries to target for prevention.	Sport-specific epidemiology.
2) Identify at risk populations.	Sport-specific epidemiology.
3) Define impact conditions resulting in head injuries from 1) and 2): - Impact dynamics	Clinical injury data paired with head impact data (video analysis, in-field measurement, reconstruction, simulation) and incident

- Impact severity - Impact opponent	circumstances (video footage, interviews, and reports).
4) Determine metric(s) related to injuries identified in 1) - Type of metric (e.g. global kinematics or simulated tissue response).	Biomechanical test data, paired with clinical injury data where necessary (cadaver, animal, volunteer or cohort head impact data, computer simulation and injury data).
5) Identify or develop a test surrogate with properties relevant to 2), 3) and 4), (e.g. mass, moment of inertia, product of inertia, surface friction, impact response, boundary conditions such as a biofidelic neck at impact)	Biomechanical data (anthropometry, human cadaver and volunteer testing). Design (e.g. computer-aided engineering), materials and manufacturing techniques.
6) Design impact test to reproduce 3), using 5). - Ensure devised test method and 4) are sensitive to changes in headgear design.	Exploratory impact testing.
7) Set effective allowable threshold for assessment metric 4).	Clinical injury data paired with biomechanical test data. Injury surveillance for monitoring set threshold effectiveness.
8) Identify minimum number of independently responding impact scenarios and severities to reduce required test configurations.	Exploratory impact testing.
9) Simplify impact test apparatus, if necessary, to be practical, repeatable and reproducible. - Impact dynamics - Impact opponent	Exploratory impact testing.
10) Draft and implement standard.	Task group participation and collaboration.

4.7 Future impact assessment of headgear in sports

In looking to the future for headgear standards, it is important not to forget the past successes of sports headgear in preventing catastrophic head injuries. A change in standards impact testing that compromises the benefits to preventing death and severe head injury achieved in many sports through current headgear standards is obviously undesirable. New test methods will likely be additions to, rather than replacements of, current test methods in certification standards unless new tests demonstrate the ability to provide at least the same level of protection as current requirements.

Certification standards play the most significant role in ensuring adequately protective headgear but they impose only a minimum level of required performance. Consumer information rating schemes are another mechanism for fostering improved headgear performance beyond meeting a standard. Rating schemes currently exist for motorcycle, bicycle, football and hockey headgear [98,126,148,264,271–274]. Further rating schemes are being proposed by researchers at Virginia Tech University for youth football, baseball and softball headgear, as well as head impact sensors [275].

The impact test methods performed under these schemes require many of the same aspects as standards tests, such as repeatability and reproducibility, and regularly develop directly from headgear research projects. Engaging stakeholders, such as headgear users, retailers, distributors and manufacturers, in what a rating scheme does and what it offers above minimum safety standards can increase the effectiveness of such schemes and drive headgear improvement [271].

5. Summary

Impact testing of headgear in current standards employs remarkably similar methods across a diverse range of sports (Table 1) even though head impact conditions can be very different. These headgear test standards have remained substantially unchanged for decades despite a shifting landscape and improved biomechanical understanding of traumatic head injuries. A great number of new impact test methods have been used by researchers in an effort to promote improved headgear design (Table 2), each allowing more headform degrees of freedom than existing standards. The diversity of these new test methods highlights the different decisions that can be made to arrive at a headgear impact test. This review provides a basis for headgear impact test development and research areas to call on for support in defining the impact condition and surrogate boundary conditions, test surrogate development and establishing assessment criteria and thresholds. Current pressing issues for continued research include definition of injurious head impact conditions in sports for which current data is sparse, establishment of reliable headgear assessment criteria related to the injuries of interest and taking into account age and size effects, and development of headgear assessment specific head and neck surrogates for at risk populations.

Acknowledgements

The authors thank C.R. Dennison, D.J. Pearsall and C. Withnall for their review of and contribution to this work.

Conflict of interest statement

Authors PAC and GPS work for a consulting company that may benefit from being associated with this work. Authors DP and PAC have interests in commercializing products related to impact injury prevention. None of the remaining parties have a conflict of interest with the submitted work.

References

[1] Pike, I., Richmond, S., Rothman, L., and Macpherson, A., eds., 2015, *Canadian Injury Prevention Resource: An Evidence-Informed Guide to Injury Prevention in Canada*, Parachute, Toronto, ON.

[2] Hoshizaki, T. B., Post, A., Oeur, R. A., and Brien, S. E., 2014, "Current and Future Concepts in Helmet and Sports Injury Prevention:," Neurosurgery, **75**, pp. S136–S148.

[3] Becker, E., 1998, "Helmet Development and Standards," *Frontiers in Head and Neck Trauma: Clinical and Biomechanical*, N. Yoganandan, F. Pintar, S. Larson, and A. Sances Jr, eds., IOS Press, Amsterdam, The Netherlands.

[4] McIntosh, A. S., Andersen, T. E., Bahr, R., Greenwald, R., Kleiven, S., Turner, M., Varese, M., and McCrory, P., 2011, "Sports Helmets Now and in the Future.," Br. J. Sports Med., **45**(16), pp. 1258–65.

[5] Mcintosh, A. S., and McCrory, P., 2005, "Preventing Head and Neck Injury," Br. J. Sports Med., **39**, pp. 314–318.

[6] Haider, A. H., Saleem, T., Bilaniuk, J. W., and Barraco, R. D., 2012, "An Evidence-Based Review: Efficacy of Safety Helmets in the Reduction of Head Injuries in Recreational Skiers and Snowboarders," J. Trauma Acute Care Surg., **73**(5), pp. 1340–1347.

[7] Sone, J. Y., Kondziolka, D., Huang, J. H., and Samadani, U., 2017, "Helmet Efficacy against Concussion and Traumatic Brain Injury: A Review," J. Neurosurg., **126**, pp. 768–781.

[8] Gurdjian, E. ., Hodgson, V. R., Hardy W.G., Patrick, L. M., and Lissner, H. R., 1964, "Evaluation of the Protective Characteristics of Helmets in Sports.," J. Trauma Inj. Infect. Crit. Care, **4**, pp. 309–24.

[9] Daneshvar, D. H., Nowinski, C. J., McKee, A. C., and Cantu, R. C., 2011, "The Epidemiology of Sport-Related Concussion," Clin. Sports Med., **30**(1), pp. 1–17.

[10] Benson, B. W., Hamilton, G. M., Meeuwisse, W. H., Mccrory, P., and Dvorak, J., 2009, "Is Protective Equipment Useful in Preventing Concussion? A Systematic Review of the Literature," Br. J. Sports Med., **43**(Suppl 1), pp. i56-67.

[11] Clay, M. B., Glover, K. L., and Lowe, D. T., 2013, "Epidemiology of Concussion in Sport: A Literature Review," J. Chiropr. Med., **12**(4), pp. 230–251.

[12] Bonfield, C. M., Shin, S. S., and Kanter, A. S., 2015, "Helmets, Head Injury and Concussion in Sport," Phys. Sportsmed., **43**(3), pp. 236–246.

[13] Daneshvar, D. H., Baugh, C. M., Nowinski, C. J., McKee, A. C., Stern, R. A., and Cantu, R. C., 2011, "Helmets and Mouth Guards: The Role of Personal Equipment in Preventing Sport-Related Concussions," Clin. Sports Med., **30**(1), pp. 145–163.

[14] Levy, M. L., Ozgur, B. M., Berry, C., Aryan, H. E., and Apuzzo, M. L. J., 2004, "Birth and Evolution of the Football Helmet," Neurosurgery, **55**(3), pp. 656–661.

[15] Odelgard, B., 1989, "The Development of Head, Face, and Neck Protectors for Ice Hockey Players," *Safety in Ice Hockey*, C. Castaldi, and E. Hoerner, eds., ASTM International, West Conshohocken, PA, pp. 220–234.

[16] British Standards Institution, 1960, *Specification for Protective Helmets for Racing Motorcyclists*, BS 1869:1960, London, England.

[17] Emery, C., and Meeuwisse, W. H., 2006, "Injury Rates, Risk Factors, and Mechanisms of Injury in Minor Hockey," Am. J. Sports Med., **34**(12), pp. 1960–1969.

[18] Gerberich, S., Finke, R., Madden, M., Priest, J., Aamoth, G., and Murray, K., 1987, "An Epidemiological Study of High School Ice Hockey Injuries," Childs Nerv. Syst. ChNS Off. J. Int. Soc. Pediatr. Neurosurg., **3**, pp. 59–64.

[19] Hutchison, M. G., 2011, *Concussions in the National Hockey League (NHL): The Video Analysis Project*, (PhD Thesis) University of Toronto.

[20] Gennarelli, T. A., 1982, "Cerebral Concussion and Diffuse Brain Injuries," *Head Injury*, P. Cooper, ed., Williams and Wilkins, Baltimore/London, pp. 83–88.

[21] Gennarelli, T. A., Thibault, L. E., Adams, J. H., Graham, D. I., Thompson, C. J., and Marcincin, R. P., 1982, "Diffuse Axonal Injury and Traumatic Coma in the Primate," Ann. Neurol., **12**, pp. 564–574.

[22] Gennarelli, T. A., and Thibault, L. E., 1982, "Biomechanics of Acute Subdural Hematoma," J. Trauma, **22**(8), pp. 680–686.
[23] Ommaya, A. K., and Gennarelli, T. A., 1974, "Cerebral Concussion and Traumatic Unconsciousness: Correlation of Experimental and Clinical Observations on Blunt Head Injuries," Brain, **97**(4), pp. 633–654.
[24] Ommaya, A. K., Hirsch, A. E., Flamm, E. S., and Mahone, R. H., 1966, "Cerebral Concussion in the Monkey: An Experimental Model," Science, **153**(3732), pp. 211–212.
[25] Margulies, S. S., and Thibault, L. E., 1992, "A Proposed Tolerance Criterion for Diffuse Axonal Injury in Man," J. Biomech., **25**(8), pp. 917–923.
[26] Depreitere, B., Van Lierde, C., Sloten, J. V., Van Audekercke, R., Van Der Perre, G., Plets, C., and Goffin, J., 2006, "Mechanics of Acute Subdural Hematomas Resulting from Bridging Vein Rupture," J. Neurosurg., **104**(6), pp. 950–956.
[27] Browne, K. D., Chen, X.-H., Meaney, D. F., and Smith, D. H., 2011, "Mild Traumatic Brain Injury and Diffuse Axonal Injury in Swine," J. Neurotrauma, **28**(9), pp. 1747–1755.
[28] Aare, M., and Halldin, P., 2003, "A New Laboratory Rig for Evaluating Helmets Subject to Oblique Impacts," Traffic Inj. Prev., **4**(3), pp. 240–248.
[29] Pang, T. Y., Thai, K. T., McIntosh, A. S., Grzebieta, R., Schilter, E., Dal Nevo, R., and Rechnitzer, G., 2011, "Head and Neck Responses in Oblique Motorcycle Helmet Impacts: A Novel Laboratory Test Method," Int. J. Crashworthiness, **16**(3), pp. 297–307.
[30] Mills, N. J., and Gilchrist, A., 1996, "Response of Helmets in Direct and Oblique Impacts," Int. J. Crashworthiness, **2**(1), pp. 7–24.
[31] McIntosh, A. S., Lai, A., and Schilter, E., 2013, "Bicycle Helmets: Head Impact Dynamics in Helmeted and Unhelmeted Oblique Impact Tests," Traffic Inj. Prev., **14**(5), pp. 501–508.
[32] McIntosh, A. S., and Lai, A., 2013, "Motorcycle Helmets: Head and Neck Dynamics in Helmeted and Unhelmeted Oblique Impacts," Traffic Inj. Prev., **14**(8), pp. 835–844.
[33] Aiello, M., Galvanetto, U., and Iannucci, L., 2007, "Numerical Simulations of Motorcycle Helmet Impact Tests," Int. J. Crashworthiness, **12**(1), pp. 1–7.
[34] Caccese, V., Ferguson, J., Lloyd, J., Edgecomb, M., Seidi, M., and Hajiaghamemar, M., 2016, "Response of an Impact Test Apparatus for Fall Protective Headgear Testing Using a Hybrid-III Head/Neck Assembly," Exp. Tech., **40**(1), pp. 413–427.
[35] Halldin, P., Gilchrist, A., and Mills, N. J., 2001, "A New Oblique Impact Test for Motorcycle Helmets," Int. J. Crashworthiness, **6**(1), pp. 53–64.
[36] Finan, J. D., Nightingale, R. W., and Myers, B. S., 2008, "The Influence of Reduced Friction on Head Injury Metrics in Helmeted Head Impacts.," Traffic Inj. Prev., **9**(5), pp. 483–8.
[37] Hansen, K., Dau, N., Feist, F., Deck, C., Willinger, R., Madey, S. M., and Bottlang, M., 2013, "Angular Impact Mitigation System for Bicycle Helmets to Reduce Head Acceleration and Risk of Traumatic Brain Injury," Accid. Anal. Prev., **59**, pp. 109–117.
[38] Mellor, A., and StClair, V., 2005, "Advanced Motorcycle Helmets," *19th International Technical Conference of the Enhanced Safety of Vehicles Proceedings*, Washington, DC, pp. 1–13.
[39] Roseveare, A. J., Plant, D. J., and Ghajari, M., 2016, "A New Helmet-Liner Design for Improved Survivability," *IRCOBI Conference Proceedings*, Malaga, Spain.
[40] Canadian Standards Association (CSA), 2009, *Cycling Helmets*, CSA Group, CAN/CSA-D113.2-M89. A National Standard of Canada.
[41] Consumer Product Safety Commission (CPSC), 1998, *Safety Standard for Bicycle Helmets*, Consumer Product Safety Commission, 16 CFR Part 1203. Washington, D.C.
[42] American National Standards Institute (ANSI), 1998, *American National Standard for Protective Headgear - for Bicyclists*, American National Standards Institute, Inc., ANSI Z90-4-1984. 16 CFR 1203-53(a)(1).New York, NY.
[43] Snell Memorial Foundation, 1998, *1995 Standard for Protective Headgear for Use in Bicycling*, Snell Memorial Foundation, Inc., North Highlands, CA.
[44] Snell Memorial Foundation, 1998, *1998 Augmentation to the 1990 Standard for Protective Headgear for Use in Bicycling*, Snell Memorial Foundation, Inc., North Highlands, CA.
[45] Snell Memorial Foundation, 1998, *1994 Standard for Protective Headgear for Use in Non-Motorized Sports*, Snell Memorial Foundation, Inc., North Highlands, CA.

[46] ASTM International, 2012, *Standard Specification for Helmets Used in Recreational Bicycling or Roller Skating*, ASTM International, F1447-12. West Conshohocken, PA.
[47] ASTM International, 2016, *Standard Specification for Helmets Used for BMX Cycling*, ASTM International, F2032-15. West Conshohocken, PA.
[48] ASTM International, 2015, *Standard Specification for Helmets Used for Downhill Mountain Bicycle Racing*, ASTM International, F1952-15. West Conshohocken, PA.
[49] European Committee for Standardization, 2014, *Helmets for Pedal Cyclists and for Users of Skateboards and Roller Skates*, European Committee for Standardization, DIN EN 1078:2014-04.
[50] ASTM International, 2015, *Standard Specification for Helmets for Non-Motorized Wheeled Vehicles Used by Infants and Toddlers*, ASTM International, F1898-15. West Conshohocken, PA.
[51] ASTM International, 2015, *Standard Specification for Helmets Used in Skateboarding and Trick Roller Skating*, ASTM International, F1492-15. West Conshohocken, PA.
[52] Canadian Standards Association (CSA), 2015, *Ice Hockey Helmets*, CSA Group, Z262.1-15. Toronto, ON, Canada.
[53] National Operating Committee on Standards for Athletic Equipment (NOCSAE), 2016, *Standard Performance Specification for Newly Manufactured Ice Hockey Helmets*, NOCSAE, ND030-11m16.
[54] ASTM International, 2015, *Standard Performance Specification for Ice Hockey Helmets*, ASTM International, F1045-15. West Conshohocken, PA.
[55] The International Organization for Standardization, 2016, *Protective Equipment for Use in Ice Hockey - Part 2: Head Protection for Skaters*, The International Organisation for Standardization, ISO 10256-2:2016(E).
[56] Canadian Standards Association (CSA), 2015, *Recreational Alpine Skiing and Snowboarding Helmets*, CSA Group, Z263.1-14. Toronto, ON.
[57] Snell Memorial Foundation, 1998, *1998 Standard for Protective Headgear for Use in Recreational Skiing and Snowboarding*, Snell Memorial Foundation, Inc., North Highlands, CA.
[58] Snell Memorial Foundation, 1998, *1998 Standard for Protective Headgear for Skiing and Other Winter Activities*, Snell Memorial Foundation, Inc., North Highlands, CA.
[59] ASTM International, 2011, *Standard Specification for Helmets Used for Recreational Snow Sports*, ASTM International, F2040-11. West Conshohocken, PA.
[60] European Committee for Standardization, 2007, *Helmets for Alpine Skiers and Snowboarders*, European Committee for Standardization, DIN EN 1077:2007-11.
[61] ASTM International, 2012, *Standard Specification for Helmets Used in Short Track Speed Ice Skating (Not to Include Hockey)*, ASTM International, F1849-07. West Conshohocken, PA.
[62] National Operating Committee on Standards for Athletic Equipment (NOCSAE), 2015, *Standard Performance Specification for Newly Manufactured Lacrosse Helmets with Faceguard*, NOCSAE, ND041-11m15.
[63] National Operating Committee on Standards for Athletic Equipment (NOCSAE), 2015, *Standard Performance Specification for Newly Manufactured Football Helmets*, NOCSAE, ND002-13m15.
[64] ASTM International, 2010, *Standard Specification for Football Helmets*, ASTM International, F717-10. West Conshohocken, PA.
[65] National Operating Committee on Standards for Athletic Equipment (NOCSAE), 2015, *Standard Performance Specification for Newly Manufactured Baseball/Softball Batter's Helmets*, NOCSAE, ND022-10m15.
[66] National Operating Committee on Standards for Athletic Equipment (NOCSAE), 2015, *Standard Performance Specification for Newly Manufactured Baseball/Softball Catcher's Helmets with Faceguard*, NOCSAE, ND024-11m15.
[67] National Operating Committee on Standards for Athletic Equipment (NOCSAE), 2015, *Standard Performance Specification for Newly Manufactured Baseball/Softball Fielder's Headgear*, NOCSAE, ND029-12m15a.

[68] Snell Memorial Foundation, 2016, *2016 Standard for Protective Headgear for Use in Horseback Ridiing*, Snell Memorial Foundation, Inc., North Highlands, CA.
[69] Snell Memorial Foundation, 2000, *2000 Standard for Protective Headgear for Use in Harness Racing*, Snell Memorial Foundation, Inc., North Highlands, CA.
[70] ASTM International, 2015, *Standard Specification for Protective Headgear Used in Horse Sports and Horseback Riding*, ASTM International, F1163-15. West Conshohocken, PA.
[71] National Operating Committee on Standards for Athletic Equipment (NOCSAE), 2015, *Standard Performance Specification for Newly Manufactured Polo Helmets*, NOCSAE, ND050-11m15.
[72] ASTM International, 2013, *Standard Specification for Protective Headgear with Faceguard Used in Bull Riding*, ASTM International, F2530-13. West Conshohocken, PA.
[73] ASTM International, 2016, *Standard Specification for Helmets Used in Pole Vaulting*, ASTM International, F2400-16. West Conshohocken, PA.
[74] ASTM International, 2015, *Standard Specification for Protective Headgear Used in Martial Arts*, ASTM International, F2397-07. West Conshohocken, PA.
[75] ASTM International, 2016, *Standard Specification for Headgear Used in Soccer*, ASTM International, F2439-06. West Conshohocken, PA.
[76] ASTM International, 2014, *Standard Specification for Testing Off-Road Motorcycle and ATV Helmets*, ASTM International, F3103-14. West Conshohocken, PA.
[77] National Highway Traffic Safety Administration, 1997, *Federal Motor Vehicle Safety Standards: Standard No. 218*, U.S. Government Printing Office.
[78] Snell Memorial Foundation, 2015, *2015 Standard for Protective Headgear for Use with Motorcycles and Other Motorized Vehicles*, Snell Memorial Foundation, Inc., North Highlands, CA.
[79] Snell Memorial Foundation, 2015, *2015 Standard for Protective Headgear for Use in Competitive Automotive Sports*, Snell Memorial Foundation, Inc., North Highlands, CA.
[80] Snell Memorial Foundation, 2016, *2016 Standard for Protective Headgear for Use in Elite Automotive Sports*, Snell Memorial Foundation, Inc., North Highlands, CA.
[81] Snell Memorial Foundation, 2015, *2015 Standard for Protective Headgear for Use in Kart Racing*, Snell Memorial Foundation, Inc., North Highlands, CA.
[82] Snell Memorial Foundation, 2016, *2016 Standard for Protective Headgear for Use in Children's Motor Sport Activities*, Snell Memorial Foundation, Inc., North Highlands, CA.
[83] Halstead, P. D., Alexander, C. F., Cook, E. M., and Drew, R. C., 2000, "Hockey Headgear and the Adequacy of Current Designs and Standards," *Safety in Ice Hockey: Third Volume*, ASTM International.
[84] The International Organization for Standardization, 1983, *Headforms for Use in the Testing of Protective Helmets*, ISO DIS 6220, Geneva, Switzerland.
[85] European Committee for Standardization, 2006, *Headforms for Use in the Testing of Protective Helmets*, European Committee for Standardization, EN 960:2006 (E).
[86] National Operating Committee on Standards for Athletic Equipment (NOCSAE), 2013, *Standard Test Method and Equipment Used in Evaluating the Performance Characteristics of Protective Headgear/Equipment*, NOCSAE, ND001-11m12.
[87] National Highway Traffic Safety Administration, 2011, *Anthropomorphic Test Devices. Subpart E Hybrid III Test Dummy. 572.32 Head.*, U.S. Government Printing Office.
[88] ASTM International, 2015, *Standard Test Methods for Equipment and Procedures Used in Evaluating the Performance Charactertistics of Protective Headgear*, ASTM International, F1446-15b. West Conshohocken, PA.
[89] Gadd, C., 1966, "Use of a Weighted-Impulse Criterion for Estimating Injury Hazard," *Proceedings of the 10th Stapp Car Crash Conference*, New York, NY, pp. 164–174.
[90] Gibson, T., Shewchenko, N., and Whyte, T., 2017, "The Skull and Brain. Test Methods, behind Armour Blunt Trauma and Helmet Design," *Military Injury Biomechanics. The Cause and Prevention of Impact Injuries*, M. Franklyn, and P. Vee Sin Lee, eds., CRC Press, Boca Raton.
[91] Harrison, T. I., Mills, N. J., and Turner, M. S., 1996, "Jockeys' Head Injuries and Skull Cap Performance," *IRCOBI Conference Proceedings*, Dublin, Ireland.

[92] Mills, N. J., and Gilchrist, A., 2008, "Oblique Impact Testing of Bicycle Helmets," Int. J. Impact Eng., **35**(9), pp. 1075–1086.
[93] Chinn, B., Bertrand, C., Siegfried, D., Doyle, D., Dietmar, O., Schuller, E., and Willinger, R., 2001, *COST 327 Motorcycle Safety Helmets*, European Commission Directorate General for Energy and Transport.
[94] Ghajari, M., Peldschus, S., Galvanetto, U., and Iannucci, L., 2013, "Effects of the Presence of the Body in Helmet Oblique Impacts," Accid. Anal. Prev., **50**, pp. 263–271.
[95] Willinger, R., Deck, C., Halldin, P., and Otte, D., 2014, "Towards Advanced Bicycle Helmet Test Methods," *International Cycling Safety Conference*, Goteborg, Sweden, pp. 18–19.
[96] Willinger, R., Halldin, P., Bogerd, C. ., Deck, C., and Fahlstedt, M., 2015, *Final Report of Working Group 3: Impact Engineering. A COST Action TU1101 / HOPE Collaboration*, COST Action TU1101 / HOPE.
[97] Bourdet, N., Mojumder, S., Piantini, S., Deck, C., Pierini, M., and Willinger, R., 2016, "Proposal of a New Motorcycle Helmet Test Method for Tangential Impact," *IRCOBI Conference Proceedings*, Malaga, Spain.
[98] Bland, M. L., McNally, C., and Rowson, S., 2018, *STAR Methodology for Bicycle Helmets*, Virginia Tech, Virginia Tech Helmet Lab.
[99] Ebrahimi, I., Golnaraghi, F., and Wang, G. G., 2015, "Factors Influencing the Oblique Impact Test of Motorcycle Helmets," Traffic Inj. Prev., **16**(4), pp. 404–408.
[100] Klug, C., Feist, F., and Tomasch, E., 2015, "Testing of Bicycle Helmets for Preadolescents," *IRCOBI Conference Proceedings*, Lyon, France, pp. 136–155.
[101] Stigson, H., Rizzi, M., Ydenius, A., Engström, E., and Kullgren, A., 2017, "Consumer Testing of Bicycle Helmets," *IRCOBI Conference Proceedings*, Antwerp, Belgium.
[102] Stuart, C. A., and Cripton, P. A., 2017, "Design of a Novel Helmet Testing Appartus Representative of Snow Sports Head Injury," *IRCOBI Conference Proceedings*, Antwerp, Belgium.
[103] Saczalski, K. J., West, M. N., Saczalski, T. K., Burton, J. L., and Pozzi, M. C., 2016, "Football Helmet Energy Absorption Degradation and Impact Performance Resulting From High Humidity and Temperature," *ASME 2016 International Mechanical Engineering Congress and Exposition*, American Society of Mechanical Engineers, p. V003T04A067–V003T04A067.
[104] Dressler, D., Richards, D., Bates, E., Van Toen, C., and Cripton, P., 2012, "Head and Neck Injury Potential With and Without Helmets During Head-First Impacts on Snow," *Skiing Trauma and Safety: 19th Volume*, R.J. Johnson, J.E. Shealy, R.M. Greenwald, and I.S. Scher, eds., ASTM International, 100 Barr Harbor Drive, PO Box C700, West Conshohocken, PA 19428-2959, pp. 235–249.
[105] Dressler, D. M., Dennison, C. R., and Cripton, P. A., 2014, "Development of an Advanced Football Helmet to Provide Increased Protection against Concussion," *Mechanism of Concussion in Sports, STP 1552*, A. Ashare, and M. Ziejewski, eds., ASTM International, 100 Barr Harbor Drive, PO Box C700, West Conshohocken, PA 19428-2959, pp. 1–18.
[106] Dressler, D. M., Dennison, C. R., Whyte, T., and Cripton, P. A., 2018, "A Novel Helmet-Mounted Device for Reducing the Potential of Catastrophic Cervical Spine Fractures and Spinal Cord Injuries in Head-First Impacts," Clin. Biomech.
[107] Ferguson, J. R., Castle, S., Harada, N., and Caccese, V., 2015, "Customized Manufacture of Protective Headgear for Primary Prevention of Fall Related Traumatic Brain Injuries," *Customized Manufacture of Protective Headgear for Primary Prevention of Fall Related Traumatic Brain Injuries*, National Harbor, MD.
[108] Seidi, M., Hajiaghamemar, M., Ferguson, J., and Caccese, V., 2015, "Injury Mitigation Performance of a Head Protection Wear with Polyurethane Honeycomb," *SAE Technical Paper 2015-01-1443*.
[109] Pellman, E. J., Viano, D. C., Tucker, A. M., Casson, I. R., Waeckerle, J. F., Maroon, J. C., Lovell, M. R., Collins, M. W., Kelly, D. F., Valadka, A. B., Cantu, R. C., Bailes, J. E., and Levy, M. L., 2003, "Concussion in Professional Football: Reconstruction of Game Impacts and Injuries," Neurosurgery, **53**(4), pp. 799–814.
[110] Viano, D. C., and Pellman, E. J., 2005, "Concussion in Professional Football: Biomechanics of the Striking Player--Part 8," Neurosurgery, **56**(2), pp. 266–280; discussion 266-280.

[111] Viano, D. C., Pellman, E. J., Withnall, C., and Shewchenko, N., 2006, "Concussion in Professional Football: Performance of Newer Helmets in Reconstructed Game Impacts - Part 13," Neurosurgery, **59**(3), pp. 591–606.
[112] Withnall, C., 2005, "Effectiveness of Headgear in Football," Br. J. Sports Med., **39**(Supplement 1), pp. i40–i48.
[113] Newman, J. A., Beusenberg, M. C., Shewchenko, N., Withnall, C., and Fournier, E., 2005, "Verification of Biomechanical Methods Employed in a Comprehensive Study of Mild Traumatic Brain Injury and the Effectiveness of American Football Helmets," J. Biomech., **38**(7), pp. 1469–1481.
[114] Butz, R. C., Knowles, B. M., Newman, J. A., and Dennison, C. R., 2015, "Effects of External Helmet Accessories on Biomechanical Measures of Head Injury Risk: An ATD Study Using the Hybrid III Headform," J. Biomech., **48**(14), pp. 3816–3824.
[115] Knowles, B. M., and Dennison, C. R., 2017, "Predicting Cumulative and Maximum Brain Strain Measures From Hybrid III Head Kinematics: A Combined Laboratory Study and Post-Hoc Regression Analysis," Ann. Biomed. Eng., **45**(9), pp. 2146–2158.
[116] Clark, J. M., Post, A., Hoshizaki, T. B., and Gilchrist, M. D., 2015, "Determining the Relationship between Linear and Rotational Acceleration and MPS for Different Magnitudes of Classified Brain Injury Risk in Ice Hockey," *IRCOBI Conference Proceedings*, Lyon, France, pp. 168–179.
[117] Clark, J. M., Post, A., Hoshizaki, T. B., and Gilchrist, M. D., 2016, "Protective Capacity of Ice Hockey Helmets against Different Impact Events," Ann. Biomed. Eng., **44**(12), pp. 3693–3704.
[118] Karton, C., Rousseau, P., Vassilyadi, M., and Hoshizaki, T. B., 2012, "The Evaluation of Speed Skating Helmet Performance through Peak Linear and Rotational Accelerations," Br. J. Sports Med., **48**(1), pp. 46–50.
[119] Posey, R., 2006, *Soccer Headgear Effects on Impact Profiles During a Heading Scenario*, (Master's Thesis) Clemson University.
[120] Aldman, B., Lundell, B., and Thorngren, L., 1976, "Non-Perpendicular Impacts: An Experimental Study on Crash Helmets," *IRCOBI Conference Proceedings*, Amsterdam, The Netherlands.
[121] Aldman, B., Lundell, B., and Thorngren, L., 1978, "Oblique Impacts - A Parametric Study in Crash Helmets," *IRCOBI Conference Proceedings*, Lyon, France.
[122] Dau, N., Hansen, K., Madey, S., and Bottlang, M., 2012, "Bicycle Helmet Impact Test Method Designed to Induce Rotational Acceleration," *Northwest Biomechanics Symposium*, Eugene, Oregon.
[123] Giacomazzi, A., Smith, T., and Kersey, R., 2009, "Analysis of the Impact Performance of Ice Hockey Helmets Using Two Different Test Methodologies," J. ASTM Int., **6**(4).
[124] Jadischke, R., Viano, D. C., McCarthy, J., and King, A. I., 2016, "The Effects of Helmet Weight on Hybrid III Head and Neck Responses by Comparing Unhelmeted and Helmeted Impacts," J. Biomech. Eng., **138**, pp. 101008–1.
[125] Pellman, E. J., Viano, D. C., Withnall, C., Shewchenko, N., Bir, C. A., and Halstead, P. D., 2006, "Concussion in Professional Football: Helmet Testing to Assess Impact Performance - Part 11," Neurosurgery, **58**(1), pp. 78–95.
[126] Rowson, B., Rowson, S., and Duma, S. M., 2015, "Hockey STAR: A Methodology for Assessing the Biomechanical Performance of Hockey Helmets," Ann. Biomed. Eng., **43**(10), pp. 2429–2443.
[127] Tyson, A. M., and Rowson, S., 2018, *Adult Football STAR Methodology*, Virginia Tech, Virginia Tech Helmet Lab.
[128] Bartsch, A., Benzel, E., Miele, V., and Prakash, V., 2012, "Impact Test Comparisons of 20th and 21st Century American Football Helmets," J. Neurosurg., **116**(1), pp. 222–233.
[129] Gwin, J. T., Chu, J. J., Diamond, S. G., Halstead, P. D., Crisco, J. J., and Greenwald, R. M., 2010, "An Investigation of the NOCSAE Linear Impactor Test Method Based on In Vivo Measures of Head Impact Acceleration in American Football," J. Biomech. Eng., **132**(1), pp. 011006–9.

[130] Rousseau, P., and Hoshizaki, T. B., 2009, "The Influence of Deflection and Neck Compliance on the Impact Dynamics of a Hybrid III Headform," Proc. Inst. Mech. Eng. Part P J. Sports Eng. Technol., **223**(3), pp. 89–97.
[131] Rousseau, P., Post, A., and Hoshizaki, T. B., 2009, "The Effects of Impact Management Materials in Ice Hockey Helmets on Head Injury Criteria," Proc. Inst. Mech. Eng. Part P J. Sports Eng. Technol., **223**(4), pp. 159–165.
[132] Rousseau, P., Post, A., and Hoshizaki, T., 2009, "A Comparison of Peak Linear and Angular Headform Accelerations Using Ice Hockey Helmets," *Fifth International Symposium on Safety in Ice Hockey*, ASTM International.
[133] Walsh, E. S., Post, A., Rousseau, P., Kendall, M., Karton, C., Oeur, A., Foreman, S., and Hoshizaki, T. B., 2012, "Dynamic Impact Response Characteristics of a Helmeted Hybrid III Headform Using a Centric and Non-Centric Impact Protocol," Proc. Inst. Mech. Eng. Part P J. Sports Eng. Technol., **226**(3–4), pp. 220–225.
[134] Post, A., Oeur, A., Hoshizaki, B., and Gilchrist, M. D., 2013, "Examination of the Relationship between Peak Linear and Angular Accelerations to Brain Deformation Metrics in Hockey Helmet Impacts," Comput. Methods Biomech. Biomed. Engin., **16**(5), pp. 511–519.
[135] Post, A., Karton, C., Hoshizaki, T. B., and Gilchrist, M. D., 2014, "Analysis of the Protective Capacity of Ice Hockey Helmets in a Concussion Injury Reconstruction," *IRCOBI Conference Proceedings*, Berlin, Germany, pp. 72–80.
[136] Post, A., Oeur, A., Hoshizaki, T. B., and Gilchrist, M. D., 2012, "The Influence of Centric and Non-Centric Impacts to American Football Helmets on the Correlation between Commonly Used Metrics in Brain Injury Research," *IRCOBI Conference Proceedings*, Dublin, Ireland, pp. 10–14.
[137] Hoshizaki, T. B., Karton, C., Oeur, R. A., Kendall, M., Dawson, L., and Post, A., 2016, "An Examination of the Current National Operating Committee on Standards for Athletic Equipment System and a New Pneumatic Ram Method for Evaluating American Football Helmet Performance to Reduce Risk of Concussion," Proc. Inst. Mech. Eng. Part P J. Sports Eng. Technol., **231**(2), pp. 83–90.
[138] Beckwith, J. G., Greenwald, R. M., and Chu, J. J., 2012, "Measuring Head Kinematics in Football: Correlation Between the Head Impact Telemetry System and Hybrid III Headform," Ann. Biomed. Eng., **40**(1), pp. 237–248.
[139] Viano, D. C., Withnall, C., and Halstead, D., 2012, "Impact Performance of Modern Football Helmets," Ann. Biomed. Eng., **40**(1), pp. 160–174.
[140] Funk, J., Crandall, J., Wonnacott, M., Withnall, C., and Drive, D. R., 2017, *NFL Linear Impactor Helmet Test Protocol*, Biomechanics Consulting & Research, LLC (Biocore).
[141] Rowson, S., McNeely, D. E., Brolinson, P. G., and Duma, S. M., 2008, "Biomechanical Analysis of Football Neck Collars," Clin. J. Sport Med., **18**(4), pp. 316–321.
[142] Viano, D. C., Withnall, C., and Wonnacott, M., 2012, "Effect of Mouthguards on Head Responses and Mandible Forces in Football Helmet Impacts," Ann. Biomed. Eng., **40**(1), pp. 47–69.
[143] Johnston, J. M., Ning, H., Kim, J.-E., Kim, Y.-H., Soni, B., Reynolds, R., Cooper, L., Andrews, J. B., and Vaidya, U., 2015, "Simulation, Fabrication and Impact Testing of a Novel Football Helmet Padding System That Decreases Rotational Acceleration," Sports Eng., **18**(1), pp. 11–20.
[144] Kis, M., Saunders, F. W., Kis Sr, M., Irrcher, I., Tator, C. H., Bishop, P. J., and Martin, W., 2013, "A Method of Evaluating Helmet Rotational Acceleration Protection Using the Kingston Impact Simulator (KIS Unit)," Clin. J. Sport Med., **23**(6), pp. 470–477.
[145] McIntosh, A. S., and Patton, D. A., 2015, "Boxing Headguard Performance in Punch Machine Tests," Br. J. Sports Med., **49**(17), pp. 1108–1112.
[146] Ivarsson, J., Viano, D. C., Lövsund, P., and Parnaik, Y., 2003, "Head Kinematics in Mini-Sled Tests of Foam Padding: Relevance of Linear Responses From Free Motion Headform (FMH) Testing to Head Angular Responses," J. Biomech. Eng., **125**(4), p. 523.
[147] King, A. I., Yang, K. H., Zhang, L., Hardy, W., and Viano, D. C., 2003, "Is Head Injury Caused by Linear or Angular Acceleration?," *IRCOBI Conference Proceedings*, Lisbon, Portugal, pp. 1–12.

[148] Tyson, A. M., Press, J., and Rowson, S., 2018, *Soccer Headgear STAR Protocol*, Virginia Tech, Virginia Tech Helmet Lab.
[149] Alem, N. M., 1980, *Head Impact Test System Development*, UM-HSRI-80-72-1, Highway Safety Research Institute, Ann Arbor, MI.
[150] Bartsch, A., Benzel, E., Miele, V., Morr, D., and Prakash, V., 2012, "Hybrid III Anthropomorphic Test Device (ATD) Response to Head Impacts and Potential Implications for Athletic Headgear Testing," Accid. Anal. Prev., **48**, pp. 285–291.
[151] Bartsch, A. J., Benzel, E. C., Miele, V. J., Morr, D. R., and Prakash, V., 2012, "Boxing and Mixed Martial Arts: Preliminary Traumatic Neuromechanical Injury Risk Analyses from Laboratory Impact Dosage Data," J. Neurosurg., **116**(5), pp. 1070–1080.
[152] O'Sullivan, D. M., Fife, G. P., Pieter, W., and Shin, I., 2013, "Safety Performance Evaluation of Taekwondo Headgear," Br. J. Sports Med., **47**(7), pp. 447–451.
[153] McIntosh, A. S., and Janda, D., 2003, "Evaluation of Cricket Helmet Performance and Comparison with Baseball and Ice Hockey Helmets," Br. J. Sports Med., **37**(4), pp. 325–330.
[154] Goldsmith, W., and Kabo, J. M., 1982, "Performance of Baseball Headgear," Am. J. Sports Med., **10**(1), pp. 31–37.
[155] Viano, D. C., McCleary, J. D., Andrzejak, D. V., and Janda, D. H., 1993, "Analysis and Comparison of Head Impacts Using Baseballs of Various Hardness and a Hybrid III Dummy," Clin. J. Sport Med., **3**, pp. 217–228.
[156] Yang, N. H., Rodowicz, K. A., and Dainty, D., 2014, "Baseball Head Impacts to the Non-Helmeted and Helmeted Hydrid III ATD," *ASME 2014 International Mechanical Engineering Congress and Exposition*, American Society of Mechanical Engineers, p. V003T03A007–V003T03A007.
[157] Stepan, L. L., Scher, I. S., and Thomas, R., 2010, "Protective Capabilities of a Watersports Helmet for Boom-to-Head Impacts During Sailing," *ASME 2010 Summer Bioengineering Conference*, American Society of Mechanical Engineers, pp. 305–306.
[158] Ghajari, M., Galvanetto, U., Iannucci, L., and Willinger, R., 2011, "Influence of the Body on the Response of the Helmeted Head during Impact," Int. J. Crashworthiness, **16**(3), pp. 285–295.
[159] Hering, A. M., and Derler, S., 2000, "Motorcycle Helmet Drop Tests Using a Hybrid III Dummy," *IRCOBI Conference Proceedings*, Montpellier, France, pp. 307–20.
[160] Aldman, B., Lundell, B., Thorngren, L., and Turbell, T., 1978, "Helmet Attenuation of the Head Response in Oblique Impacts to the Ground," *IRCOBI Conference Proceedings*, Lyon, France.
[161] Corner, J. P., Whitney, C. W., O'Rourke, N., and Morgan, D. E., 1987, *Motorcycle and Bicycle Protective Helmets: Requirements Resulting from a Post Crash Study and Experimental Research*, CR 55, Federal Office of Road Safety, Canberra, ACT, Australia.
[162] Hodgson, V. R., 1990, *Impact, Skid and Retention Tests on a Representative Group of Bicycle Helmets to Determine Their Head-Neck Protective Characteristics*, Wayne State University, Department of Neurosurgery.
[163] Scher, I., Richards, D., and Carhart, M., 2006, "Head Injury in Snowboarding: Evaluating the Protective Role of Helmets," J. ASTM Int., **3**(4), pp. 1–8.
[164] Richards, D., Carhart, M., Scher, I., Thomas, R., and Hurlen, N., 2009, "Head Kinematics during Experimental Snowboard Falls: Implications for Snow Helmet Standards," J. ASTM Int., **5**(6), pp. 101–107.
[165] Scher, I., Richards, D., Carhart, M., Thomas, R., Hurlen, N., and Lam, T., 2009, "Pediatric Head and Neck Injuries in Snow Sports: Evaluating the Influence of Helmets," *Skiing Trauma and Safety: 17th Volume*, ASTM International.
[166] Yamaguchi, G. T., Dainty, D., Lam, T., and Watson, H., 2014, "Performance of Certified Climbing Helmets During Simulated Climbing Falls," J. Test. Eval., **42**(5), p. 20130122.
[167] Kimpara, H., and Iwamoto, M., 2012, "Mild Traumatic Brain Injury Predictors Based on Angular Accelerations During Impacts," Ann. Biomed. Eng., **40**(1), pp. 114–126.
[168] Kimpara, H., Nakahira, Y., Iawmoto, M., Rowson, S., and Duma, S., 2011, "Head Injury Prediction Methods Based on 6 Degree of Freedom Head Acceleration Measurements during Impact," Int. J. Automot. Eng., **2**, pp. 11–19.

[169] Newman, J. A., 1986, “A Generalised Acceleration Model for Brain Injury Threshold (GAMBIT),” *IRCOBI Conference Proceedings*, Zurich, Switzerland.
[170] Newman, J. A., Shewchenko, N., and Welbourne, E., 2000, “A Proposed New Biomechanical Head Injury Assessment Function - the Maximum Power Index.,” Stapp Car Crash J., **44**, pp. 215–47.
[171] Sahoo, D., Deck, C., and Willinger, R., 2016, “Brain Injury Tolerance Limit Based on Computation of Axonal Strain,” Accid. Anal. Prev., **92**, pp. 53–70.
[172] Takhounts, E. G., Craig, M. J., Moorhouse, K., McFadden, J., and Hasija, V., 2013, “Development of Brain Injury Criteria (BrIC),” Stapp Car Crash J., **57**, pp. 243–266.
[173] Aare, M., Kleiven, S., and Halldin, P., 2004, “Injury Tolerances for Oblique Impact Helmet Testing,” Int. J. Crashworthiness, **9**(1), pp. 15–23.
[174] Mertz, H. J., Prasad, P., and Nusholtz, G., 1996, *Head Injury Risk Assessment for Forehead Impacts*, SAE Technical Paper 960099.
[175] Ommaya, A. K., Goldsmith, W., and Thibault, L., 2002, “Biomechanics and Neuropathology of Adult and Paediatric Head Injury,” Br J Neurosurg, **16**(3), pp. 220–42.
[176] Bain, A. C., and Meaney, D. F., 1999, “Thresholds for Mechanical Injury to the In Vivo White Matter,” *43rd Stapp Car Crash Conference*, San Diego, California.
[177] Bain, A. C., and Meaney, D. F., 2000, “Tissue-Level Thresholds for Axonal Damage in an Experimental Model of Central Nervous System White Matter Injury,” J Biomech Eng, **122**(6), pp. 615–22.
[178] Morrison III, B., Cater, H. L., Wang, C. C., Thomas, F. C., and others, 2003, “A Tissue Level Tolerance Criterion for Living Brain Developed with an in Vitro Model of Traumatic Mechanical Loading,” Stapp Car Crash J., **47**, p. 93.
[179] Dressler, D., Richards, D., Bates, E., Van Toen, C., and Cripton, P., 2012, *Head and Neck Injury Potential With and Without Helmets During Head-First Impacts on Snow*, ASTM International, West Conshohocken, PA.
[180] Kleiven, S., 2007, “Predictors for Traumatic Brain Injuries Evaluated through Accident Reconstructions,” Stapp Car Crash J., **51**, pp. 81–114.
[181] Zhang, L., Yang, K. H., and King, A. I., 2004, “A Proposed Injury Threshold for Mild Traumatic Brain Injury,” J. Biomech. Eng., **126**, pp. 226–236.
[182] Takhounts, E., Hasija, V., Ridella, S. A., Rowson, S., and Duma, S. M., 2011, “Kinematic Rotational Brain Injury Criterion (BRIC),” *22nd Enhanced Safety of Vehicles Conference Proceedings*, p. Paper Number 11-0263.
[183] Giordano, C., and Kleiven, S., 2014, “Evaluation of Axonal Strain as a Predictor for Mild Traumatic Brain Injuries Using Finite Element Modeling,” Stapp Car Crash J., **58**, p. 29.
[184] Hurt, H. H., and Thom, D. R., 1994, “Accident Performance of Motorcycle and Bicycle Safety Helmets,” *Head and Neck Injuries in Sports ASTM STP 1229*, American Society for Testing and Materials, Philadelphia.
[185] Pellman, E. J., Viano, D. C., Tucker, A. M., and Casson, I. R., 2003, “Concussion in Professional Football: Location and Direction of Helmet Impacts-Part 2,” Neurosurgery, **53**(6), p. 1328.
[186] Viano, D. C., Casson, I. R., and Pellman, E. J., 2007, “Concussion in Professional Football: Biomechanics of the Struck Player--Part 14,” Neurosurgery, **61**(2), pp. 313–327; discussion 327-328.
[187] National Operating Committee on Standards for Athletic Equipment (NOCSAE), 2016, *Standard Pneumatic Ram Test Method and Equipment Used in Evaluating the Performance Characteristics of Protective Headgear and Face Guards*, NOCSAE, ND081-14m15.
[188] National Operating Committee on Standards for Athletic Equipment (NOCSAE), 2019, *Standard Performance Specification for Newly Manufactured Football Helmets*, NOCSAE, ND002-17m19.
[189] Lessley, D. J., Kent, R. W., Funk, J. R., Sherwood, C. P., Cormier, J. M., Crandall, J. R., Arbogast, K. B., and Myers, B. S., 2018, “Video Analysis of Reported Concussion Events in the National Football League During the 2015-2016 and 2016-2017 Seasons,” Am. J. Sports Med., **46**(14), pp. 3502–3510.

[190] Willinger, R., Deck, C., Halldin, P., and Otte, D., 2014, “Towards Advanced Bicycle Helmet Test Methods,” *International Cycling Safety Conference*, Gothenburg, Sweden, pp. 1–11.
[191] Bonin, S. J., Luck, J. F., Bass, C. R., Gardiner, J. C., Onar-Thomas, A., Asfour, S. S., and Siegmund, G. P., 2017, “Dynamic Response and Residual Helmet Liner Crush Using Cadaver Heads and Standard Headforms,” Ann. Biomed. Eng., **45**(3), pp. 656–667.
[192] McIntosh, A. S., and Patton, D. A., 2012, “Impact Reconstruction from Damage to Pedal and Motorcycle Helmets,” Proc. Inst. Mech. Eng. Part P J. Sports Eng. Technol., **226**(3–4), pp. 274–281.
[193] Bourdet, N., Deck, C., Carreira, R. P., and Willinger, R., 2012, “Head Impact Conditions in the Case of Cyclist Falls,” Proc. Inst. Mech. Eng. Part P J. Sports Eng. Technol., **226**(3–4), pp. 282–289.
[194] Yamazaki, J., Gilgien, M., Kleiven, S., Mcintosh, A. S., Nachbauer, W., Müller, E., Bere, T., Bahr, R., and Krosshaug, T., 2015, “Analysis of a Severe Head Injury in World Cup Alpine Skiing,” Med. Sci. Sports Exerc., **47**(6), pp. 1113–1118.
[195] Rowson, S., Duma, S. M., Beckwith, J. G., Chu, J. J., Greenwald, R. M., Crisco, J. J., Brolinson, P. G., Duhaime, A.-C., McAllister, T. W., and Maerlender, A. C., 2012, “Rotational Head Kinematics in Football Impacts: An Injury Risk Function for Concussion,” Ann. Biomed. Eng., **40**(1), pp. 1–13.
[196] Greenwald, R. M., Chu, J. J., Beckwith, J. G., Gwin, J. T., Buck, A. T., and Crisco, J. J., 2006, “On-Field Measurement of Head Impact Acceleration in Helmeted Sports,” *American Society of Biomechanics Annual Meeting*, Blacksburg, VA.
[197] Duhaime, A.-C., Beckwith, J. G., Maerlender, A. C., McAllister, T. W., Crisco, J. J., Duma, S. M., Brolinson, P. G., Rowson, S., Flashman, L. A., Chu, J. J., and Greenwald, R. M., 2012, “Spectrum of Acute Clinical Characteristics of Diagnosed Concussions in College Athletes Wearing Instrumented Helmets: Clinical Article,” J. Neurosurg., **117**(6), pp. 1092–1099.
[198] Wilcox, B. J., Beckwith, J. G., Greenwald, R. M., Chu, J. J., McAllister, T. W., Flashman, L. A., Maerlender, A. C., Duhaime, A.-C., and Crisco, J. J., 2014, “Head Impact Exposure in Male and Female Collegiate Ice Hockey Players,” J. Biomech., **47**(1), pp. 109–114.
[199] Kerr, Z. Y., Littleton, A. C., Cox, L. M., DeFreese, J. D., Varangis, E., Lynall, R. C., Schmidt, J. D., Marshall, S. W., and Guskiewicz, K. M., 2015, “Estimating Contact Exposure in Football Using the Head Impact Exposure Estimate,” J. Neurotrauma, **32**(14), pp. 1083–1089.
[200] McCuen, E., Svaldi, D., Breedlove, K., Kraz, N., Cummiskey, B., Breedlove, E. L., Traver, J., Desmond, K. F., Hannemann, R. E., Zanath, E., Guerra, A., Leverenz, L., Talavage, T. M., and Nauman, E. A., 2015, “Collegiate Women’s Soccer Players Suffer Greater Cumulative Head Impacts than Their High School Counterparts,” J. Biomech., **48**(13), pp. 3720–3723.
[201] King, D., Hecimovich, M., Clark, T., and Gissane, C., 2017, “Measurement of the Head Impacts in a Sub-Elite Australian Rules Football Team with an Instrumented Patch: An Exploratory Analysis,” Int. J. Sports Sci. Coach., **12**(3), pp. 359–370.
[202] King, D., Hume, P., Gissane, C., and Clark, T., 2017, “Head Impacts in a Junior Rugby League Team Measured with a Wireless Head Impact Sensor: An Exploratory Analysis,” J. Neurosurg. Pediatr., **19**(1), pp. 13–23.
[203] King, D. A., Hume, P. A., Gissane, C., and Clark, T. N., 2016, “Similar Head Impact Acceleration Measured Using Instrumented Ear Patches in a Junior Rugby Union Team during Matches in Comparison with Other Sports,” J. Neurosurg. Pediatr., **18**(1), pp. 65–72.
[204] O’Sullivan, D. M., and Fife, G. P., 2017, “Biomechanical Head Impact Characteristics during Sparring Practice Sessions in High School Taekwondo Athletes,” J. Neurosurg. Pediatr., **19**(6), pp. 662–667.
[205] Jadischke, R., Viano, D., Dau, N., King, A., and McCarthy, J., 2013, “On the Accuracy of the Head Impact Telemetry (HIT) System Used in Football Helmets,” J. Biomech., **46**, pp. 2310–2315.
[206] Allison, M. A., Kang, Y. S., Bolte, J. H., 4th, Maltese, M. R., and Arbogast, K. B., 2014, “Validation of a Helmet-Based System to Measure Head Impact Biomechanics in Ice Hockey,” Med. Sci. Sports Exerc., **46**(1), pp. 115–123.

[207] Siegmund, G. P., Guskiewicz, K. M., Marshall, S. W., DeMarco, A. L., and Bonin, S. J., 2016, "Laboratory Validation of Two Wearable Sensor Systems for Measuring Head Impact Severity in Football Players," Ann. Biomed. Eng., **44**(4), pp. 1257–1274.
[208] Siegmund, G., Bonin, S. J., Luck, J. F., and Bass, C. R. D., 2015, "Validation of a Skin-Mounted Sensor for Measuring in-Vivo Head Impacts," *IRCOBI Conference Proceedings*, Lyon, France.
[209] Kuo, C., Wu, L., Loza, J., Senif, D., Anderson, S. C., and Camarillo, D. B., 2018, "Comparison of Video-Based and Sensor-Based Head Impact Exposure," PLOS ONE, **13**(6), p. e0199238.
[210] Nevins, D., Hildenbrand, K., Kensrud, J., Vasavada, A., and Smith, L., 2016, "Field Evaluation of a Small Form-Factor Head Impact Sensor for Use in Soccer," Procedia Eng., **147**, pp. 186–190.
[211] Cortes, N., Lincoln, A. E., Myer, G. D., Hepburn, L., Higgins, M., Putukian, M., and Caswell, S. V., 2017, "Video Analysis Verification of Head Impact Events Measured by Wearable Sensors," Am. J. Sports Med., **45**(10), pp. 2379–2387.
[212] Lon, S., 2014, *A New Helmet Testing Method to Assess Potential Damages in the Brain and the Head Due to Rotational Energy*, (Master's Thesis) Royal Institute of Technology KTH.
[213] Post, A., Clark, J. M., Robertson, D. G. E., Hoshizaki, T. B., and Gilchrist, M. D., 2017, "The Effect of Acceleration Signal Processing for Head Impact Numeric Simulations," Sports Eng., **20**(2), pp. 111–119.
[214] Hoshizaki, B., Vassilyadi, M., Post, A., and Oeur, A., 2012, "Performance Analysis of Winter Activity Protection Headgear for Young Children: Laboratory Investigation," J. Neurosurg. Pediatr., **9**(2), pp. 133–138.
[215] King, A. I., Yang, K. H., Zhang, L., Hardy, W., and Viano, D. C., 2003, "Is Head Injury Caused by Linear or Angular Acceleration," *IRCOBI Conference Proceedings*, Lisbon, Portugal, pp. 25–26.
[216] Kendall, M., Post, A., Rousseau, P., and Hoshizaki, T. B., 2014, "The Effect of Shoulder Pad Design on Reducing Peak Resultant Linear and Rotational Acceleration in Shoulder-to-Head Impacts," *ASTM Symposium on Mechanism of Concussion in Sports*, Atlanta, GA, pp. 142–152.
[217] Becker, E. B., Anishchenko, D., and Palmer, S., 2015, "Motorcycle Helmet Impact Response at Various Levels of Severity for Different Standard Certifications," *IRCOBI Conference Proceedings*, Lyon, France, pp. 814–824.
[218] Duma, S. M., Manoogian, S. J., Bussone, W. R., Brolinson, P. G., Goforth, M. W., Donnenwerth, J. J., Greenwald, R. M., Chu, J. J., and Crisco, J. J., 2005, "Analysis of Real-Time Head Accelerations in Collegiate Football Players," Clin. J. Sport Med., **15**(1), pp. 3–8.
[219] Brolinson, P. G., Manoogian, S., McNeely, D., Goforth, M., Greenwald, R., and Duma, S., 2006, "Analysis of Linear Head Accelerations from Collegiate Football Impacts," Curr. Sports Med. Rep., **5**(1), p. 23.
[220] McAllister, T. W., Ford, J. C., Ji, S., Beckwith, J. G., Flashman, L. A., Paulsen, K., and Greenwald, R. M., 2012, "Maximum Principal Strain and Strain Rate Associated with Concussion Diagnosis Correlates with Changes in Corpus Callosum White Matter Indices," Ann. Biomed. Eng., **40**(1), pp. 127–140.
[221] DeMarco, A. L., Chimich, D. D., Gardiner, J. C., Nightingale, R. W., and Siegmund, G. P., 2010, "The Impact Response of Motorcycle Helmets at Different Impact Severities," Accid. Anal. Prev., **42**(6), pp. 1778–1784.
[222] DeMarco, A. L., Chimich, D. D., Gardiner, J. C., and Siegmund, G. P., 2016, "The Impact Response of Traditional and BMX-Style Bicycle Helmets at Different Impact Severities," Accid. Anal. Prev., **92**, pp. 175–183.
[223] Rowson, S., Daniel, R. W., and Duma, S. M., 2013, "Biomechanical Performance of Leather and Modern Football Helmets," J. Neurosurg., **119**(3), pp. 805–809.
[224] National Operating Committee on Standards for Athletic Equipment (NOCSAE), 2018, "Certification to NOCSAE Standards and Add-On Helmet Products" [Online]. Available: https://nocsae.org/wp-content/uploads/2018/05/NOCSAE-Add-On-Products-Release-5-8-18_.pdf.

[225] Conidi, F. X., 2015, “Helmets, Sensors, and More: A Review,” Pract. Neurol., **15**(2), pp. 32–36.
[226] Breedlove, K. M., Breedlove, E., Nauman, E., Bowman, T. G., and Lininger, M. R., 2017, “The Ability of an Aftermarket Helmet Add-On Device to Reduce Impact-Force Accelerations During Drop Tests,” J. Athl. Train., **52**(9), pp. 802–808.
[227] Hodgson, V. R., 1975, “National Operating Committee on Standards for Athletic Equipment Football Helmet Certification Program,” Med. Sci. Sports, **7**(3), pp. 225–32.
[228] Mertz, H. J., 1985, “Biofidelity of the Hybrid III Head,” *SAE Government Industry Meeting and Exposition*, SAE International.
[229] Loyd, A. M., Nightingale, R. W., Song, Y., Luck, J. F., Cutcliffe, H., Myers, B. S., and Bass, C., 2014, “The Response of the Adult and ATD Heads to Impacts onto a Rigid Surface,” Accid. Anal. Prev., **72**, pp. 219–229.
[230] MacAlister, A., 2013, “Surrogate Head Forms for the Evaluation of Head Injury Risk,” *Brain Injuries and Biomechanics Symposium*, Washington, DC.
[231] Cobb, B. R., MacAlister, A., Young, T. J., Kemper, A. R., Rowson, S., and Duma, S. M., 2015, “Quantitative Comparison of Hybrid III and NOCSAE Headform Shape Characteristics and Implications on Football Helmet Fit,” Proc IMechE Part P J Sports Eng. Technol., **229**, pp. 39–46.
[232] Warnica, M. J., 2015, *Biomechanical Assessment of Cycling Helmets: The Influence of Headform and Impact Velocity Based on Cycling Collisions Associated with Injury Claims*, (Master’s Thesis) University of Waterloo.
[233] Kendall, M., Walsh, E. S., and Hoshizaki, T. B., 2012, “Comparison between Hybrid III and Hodgson–WSU Headforms by Linear and Angular Dynamic Impact Response,” Proc. Inst. Mech. Eng. Part P J. Sports Eng. Technol., **226**(3–4), pp. 260–265.
[234] Willinger, R., Halldin, P., Bogerd, C. P., Deck, C., and Fahlstedt, M., 2015, *Final Report of Working Group 3: Impact Engineering*, COST Action TU1101/HOPE, Brussels, Belgium.
[235] Loyd, A. M., 2011, *Studies of the Human Head from Neonate to Adult: An Inertial, Geometrical and Structural Analysis with Comparisons to the ATD Head*, (PhD Thesis) Duke University.
[236] Trotta, A., Ní Annaidh, A., Burek, R. O., Pelgrims, B., and Ivens, J., 2018, “Evaluation of the Head-Helmet Sliding Properties in an Impact Test,” J. Biomech., **75**, pp. 28–34.
[237] Halldin, P., 2017, “CEN/TC 158 Working Group 11 - Status Report May 2017. Focus on Headforms Questions.”
[238] Yoganandan, N., Pintar, F. A., Zhang, J., and Baisden, J. L., 2009, “Physical Properties of the Human Head: Mass, Center of Gravity and Moment of Inertia,” J. Biomech., **42**(9), pp. 1177–1192.
[239] Funk, J. R., Quesada, R. E., Miles, A. M., and Crandall, J. R., 2018, “Inertial Properties of Football Helmets,” J. Biomech. Eng., **140**(6), p. 064501.
[240] Fahlstedt, M., Halldin, P., Alvarez, V. S., and Kleiven, S., 2016, “Influence of the Body and Neck on Head Kinematics and Brain Injury Risk in Bicycle Accident Situations,” *IRCOBI Conference Proceedings*, Malaga, Spain.
[241] Verschueren, P., 2009, *Biomechanical Analysis of Head Injuries Related to Bicycle Accidents and a New Bicycle Helmet Concept*, Katholieke Universiteit Leuven, (PhD Thesis) Katholieke Universiteit Leuven.
[242] Nightingale, R. W., McElhaney, J. H., Richardson, W. J., and Myers, B. S., 1996, “Dynamic Response of the Head and Cervical Spine to Axial Impact Loading,” J. Biomech., **29**(3), pp. 307–318.
[243] Forero Ruedo, M. A., 2009, *Equestrian Helmet Design: A Computational and Head Impact, Biomechanics Simulation Approach*, University College Dublin, (PhD Thesis) University College Dublin.
[244] Beusenberg, M., Shewchenko, N., Newman, J., de Lange, R., and Cappon, H., 2001, “Head, Neck, and Body Coupling in Reconstructions of Helmeted Head Impacts,” *IRCOBI Conference Proceedings*, Isle of Man, UK, pp. 295–310.

[245] Feist, F., and Klug, C., 2016, “A Numerical Study on the Influence of the Upper Body and Neck on Head Kinematics in Tangential Bicycle Helmet Impact,” *IRCOBI Conference Proceedings*, Malaga, Spain, p. 23.
[246] Foster, J., Kortge, J., and Wolanin, M., 1977, “Hybrid III - A Biomechanically Based Crash Test Dummy,” Stapp Car Crash J., **21**, pp. 975–1014.
[247] Myers, B. S., McElhaney, J. H., Doherty, B. J., Paver, J. G., Nightingale, R. W., Ladd, T. P., and Gray, L., 1989, “Response of the Human Cervical Spine to Torsion,” *The 33rd Stapp Car Crash Conference*, Washington, DC, pp. 215–222.
[248] Svensson, M. Y., and Lovsund, P., 1992, “A Dummy for Rear-End Collisions - Development and Validation of a New Dummy-Neck,” *IRCOBI Conference Proceedings*, Verona, Italy, pp. 299–310.
[249] Herbst, B., Forrest, S., and Chang, D., 1998, “Fidelity of Anthropometric Test Dummy Necks in Rollover Accidents,” *Proceedings of the 16th International Technical Conference on the Enhanced Safety of Vehicles*, Windsor, Ontario, Canada.
[250] Cripton, P., 2013, “Perspective: Protecting the Neck,” Nature, **503**(7475), p. S13.
[251] Nelson, T. S., and Cripton, P. A., 2010, “A New Biofidelic Sagittal Plane Surrogate Neck for Head-First Impacts,” Traffic Inj. Prev., **11**(3), pp. 309–319.
[252] Clark, J. M., 2015, *Evaluation of the Protective Capacity of Ice Hockey Goaltender Masks for Three Accident Events Using Dynamic Response and Brain Stress and Strain*, (Master’s Thesis) University of Ottawa.
[253] Walsh, E. S., Kendall, M., Post, A., Meehan, A., and Hoshizaki, T. B., 2018, “Comparative Analysis of Hybrid III Neckform and an Unbiased Neckform,” Sports Eng.
[254] Walsh, E. S., Kendall, M., Hoshizaki, T. B., and Gilchrist, M. D., 2014, “Dynamic Impact Response and Predicted Brain Tissue Deformation Comparisons for an Impacted Hybrid III Headform With and Without a Neckform and Torso Masses,” *IRCOBI Conference Proceedings*, Berlin, Germany.
[255] Lissner, H. R., Lebow, M., and Evans, F. G., 1960, “Experimental Studies on the Relation between Acceleration and Intracranial Pressure Changes in Man,” Surg. Gynecol. Obstet., **111**, pp. 329–338.
[256] Gurdjian, E. S., Lissner, H. R., Evans, F. G., Patrick, L. M., and Hardy, W. G., 1961, “Intracranial Pressure and Acceleration Accompanying Head Impacts in Human Cadavers,” Surg. Gynecol. Obstet., **v113**, p. 195.
[257] Prasad, P., Melvin, J. W., Huelke, D. F., King, A. I., and Nyquist, G. ., 1985, “Head,” *Review of Biomechanical Impact Response and Injury in the Automotive Environment. Task B Final Report*, National Highway Traffic Safety Administration. U.S. Department of Transport, Ann Arbor, MI.
[258] Gabler, L. F., Crandall, J. R., and Panzer, M. B., 2016, “Assessment of Kinematic Brain Injury Metrics for Predicting Strain Responses in Diverse Automotive Impact Conditions,” Ann. Biomed. Eng., **44**(12), pp. 3705–3718.
[259] Graham, R., Rivara, F. P., Ford, M. A., and Spicer, C. M., eds., 2014, *Sports-Related Concussions in Youth: Improving the Science, Changing the Culture*, The National Academies Press, Washington, D.C.
[260] Horgan, T. J., and Gilchrist, M. D., 2003, “The Creation of Three-Dimensional Finite Element Models for Simulating Head Impact Biomechanics,” Int. J. Crashworthiness, **8**(4), pp. 353–366.
[261] Deck, C., and Willinger, R., 2008, “Improved Head Injury Criteria Based on Head FE Model,” Int. J. Crashworthiness, **13**(6), pp. 667–678.
[262] Ghajari, M., Hellyer, P. J., and Sharp, D. J., 2017, “Computational Modelling of Traumatic Brain Injury Predicts the Location of Chronic Traumatic Encephalopathy Pathology,” Brain, **140**(2), pp. 333–343.
[263] Zhao, W., Kuo, C., Wu, L., Camarillo, D. B., and Ji, S., 2017, “Performance Evaluation of a Pre-Computed Brain Response Atlas in Dummy Head Impacts,” Ann. Biomed. Eng., **45**(10), pp. 2437–2450.

[264] Rowson, S., and Duma, S. M., 2011, “Development of the STAR Evaluation System for Football Helmets: Integrating Player Head Impact Exposure and Risk of Concussion,” Ann. Biomed. Eng., **39**(8), pp. 2130–2140.
[265] Camarillo, D. B., Shull, P. B., Mattson, J., Shultz, R., and Garza, D., 2013, “An Instrumented Mouthguard for Measuring Linear and Angular Head Impact Kinematics in American Football,” Ann. Biomed. Eng., **41**(9), pp. 1939–1949.
[266] Bartsch, A., Samorezov, S., Benzel, E., Miele, V., and Brett, D., “Validation of an ‘Intelligent Mouthguard’ Single Event Head Impact Dosimeter,” Stapp Car Crash J., **58**, pp. 1–27.
[267] 2018, “NFL Safety Tech, Part Four: The Future of Impact and Concussion Monitoring,” SportTechie [Online]. Available: https://www.sporttechie.com/nfl-safety-tech-part-four-concussion-mouthguard-impact-sensor/. [Accessed: 01-Feb-2019].
[268] Smith, A. M., Stuart, M. J., Roberts, W. O., Dodick, D. W., Finnoff, J. T., Jorgensen, J. K., and Krause, D. A., 2017, “Concussion in Ice Hockey: Current Gaps and Future Directions in an Objective Diagnosis,” Clin. J. Sport Med., **27**(5), pp. 503–509.
[269] Broglio, S. P., Guskiewicz, K. M., and Norwig, J., 2017, “If You’re Not Measuring, You’re Guessing: The Advent of Objective Concussion Assessments,” J. Athl. Train., **52**(3), pp. 160–166.
[270] Swart, R., 2004, “Headgear and Helmets. ASTM Subcommittee Responds to Changing Activities and Perceptions,” ASTM Stand. News.
[271] Delmonte, E., Martin, P., Helman, S., Hynd, D., Fernandez-Medina, K., and Kinnear, N., 2015, *SHARP: A Study of Its Effect on the UK Motorcycle Helmet Market*, PPR747, Transport Research Laboratory.
[272] McIntosh, A., Suratno, B., Haley, J., and Truong, J., 2013, “Consumer Rating and Assessment of Safety Helmets for Motorcyclists,” *Proceedings of the 23rd International Technical Conference on the Enchanced Safety of Vehicles*, Seoul, South Korea.
[273] Consumer Reports, 2017, *Bike Helmets*, https://www.consumerreports.org/cro/bike-helmets.htm.
[274] NFL Enterprises LLC, “Helmet Laboratory Testing Performance Results,” NFL Play Smart Play Safe [Online]. Available: https://www.playsmartplaysafe.com/resource/helmet-laboratory-testing-performance-results/. [Accessed: 01-Feb-2019].
[275] Biomedical Engineering and Mechanics, Virginia Tech, 2018, “Virginia Tech Helmet Ratings” [Online]. Available: https://www.beam.vt.edu/helmet/. [Accessed: 29-Aug-2018].

Figure Captions

Figure 1 Guided drop test apparatus used in ASTM, CSA, CPSC, ANSI and Snell standards (Reproduced, with permission from ASTM [88], copyright ASTM International, 100 Barr Harbor Drive. West Conshohocken, PA 19428).

Figure 2 Falling headform with guided carrier test apparatus used in CEN and ISO standards [55]. 1) steel base; 2) anvil; 3) guides; 4) support dolly; and 5) headform with helmet. (Copied with permission of the Standards Council of Canada (SCC) on behalf of ISO).

Table Captions

Table 1 Summary of impact energy attenuation tests in headgear standards for sports

Table 2 Review of new headgear test methods found in literature and/or at relevant conferences

Table 3 Mass and inertial properties comparison between human cadaver data and automotive ATD headforms.

Table 4 Sports headgear impact test development roadmap for certification standards.