Effectiveness of headgear in football

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SUPPLEMENT

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Objectives: Commercial headgear is currently being used by football players of all ages and skill levels to provide protection from heading and direct impact. The clinical and biomechanical effectiveness of the headgear in attenuating these types of impact is not well defined or understood. This study was conducted to determine whether football headgear has an effect on head impact responses.

Methods: Controlled laboratory tests were conducted with a human volunteer and surrogate head/neck system. The impact attenuation of three commercial headgears during ball impact speeds of 6–30 m/s and in head to head contact with a closing speed of 2–5 m/s was quantified. The human subject, instrumented to measure linear and angular head accelerations, was exposed to low severity impacts during heading in the unprotected and protected states. High severity heading contact and head to head impacts were studied with a biofidelic surrogate headform instrumented to measure linear and angular head responses. Subject and surrogate responses were compared with published injury assessment functions associated with mild traumatic brain injury (MTBI).

Results: For ball impacts, none of the headgear provided attenuation over the full range of impact speeds. Head responses with or without headgear were not significantly different ($p > 0.05$) and remained well below levels associated with MTBI. In head to head impact tests the headgear provided an overall 33% reduction in impact response.

Conclusion: The football headgear models tested did not provide benefit during ball impact. This is probably because of the large amount of ball deformation relative to headband thickness. However, the headgear provided measurable benefit during head to head impacts.

Brain injury due to heading has recently received wide attention in the media creating a concern for the safety of the sport and players. Researchers have studied the neurological, neuropsychological, and cognitive impairments in active amateur and professional players, as well as in former professional players, with mixed findings. In a series of studies of active and retired players, cognitive deficits were suggested to be associated with repeated headings although the findings were not definitive.[1–3] Cerebral computed tomography (CT) scans of former players showed cerebral atrophy and a change in ventricular and linear dimensions. Neuropsychological examinations showed mild to severe cognitive deficits in former players. The results of these studies have been referenced extensively in literature when justifying the need for football headgear.

However, the studies have been noted to be flawed due to poor methodology including lack of controls and pre-injury data, selection bias, failure to control acute injuries, and lack of blind observers.[4] The findings may also have limited applicability to modern day players since they included former professional football players who had used old generation balls. Other studies of chronic brain injury indicate a lack of correlation with heading.[5–8] Similar conclusions were drawn in a study of amateur and professional players indicating that both concussive injury and heading were associated with diminished cognitive function.[9 10] A link was found between the number of concussions and test performance on memory and planning. Another link was found between the number of headers and neuropsychological test performance in professional players. However, Kirkendall and Garrett[11] stated that these results may have been biased by selection of the control groups.

The cognitive performance of youth players (average age 11.5 years) has been studied in relation to heading.[12] No abnormalities were found with the exception of difficulty in learning new words, and 49% reported having headaches after heading. In a study of active collegiate players and control groups, Guskiewicz et al[8] showed no evidence of diminished neuropsychological performance.

From the available research, it is not known if there is a relation between sub-concussive headings and chronic cognitive impairment.[4] Further, it has been stated that head to ball contact is unlikely to cause chronic neurological injury, and that although concussion can occur from head to head impacts it is uncommon and unlikely to contribute to cumulative injuries. Direct impact to the head can cause concussive injuries with relatively severe outcomes including death. Acute head injuries have also been suggested as a cause of long-term neuropsychological changes.[13]

Head injuries resulting from impacts during head to head, head to ground, head to goal post, and head to body extremity contacts have also been reported.[13] These tend to result in more severe impacts which lead to lacerations, mild traumatic brain injury, and traumatic brain injury. The contribution of acute impacts to chronic injuries was noted to be unlikely.[4] In a study of 29 concussions in elite college players over a two year period, the mechanisms of injury were identified.[13] The commonest causes were player to player contact, unexpected head to ball impacts from close range and, head to elbow. No injuries due to voluntary heading were reported. The distribution of mechanisms causing concussion is shown in fig 1.

The Consumer Product Safety Commission in the USA[14] reported that head to player contact is the most prevalent cause of all reported head injuries. Distribution of the mechanisms for all types of injury in this study is shown in fig 2.

Findings similar to the previous two studies were also found in research focused on repeated trauma to the brain in football.[3] Head trauma in active and former Norwegian players was caused partly by headings, head to head contact, falls to the ground, and other contacts. A later study
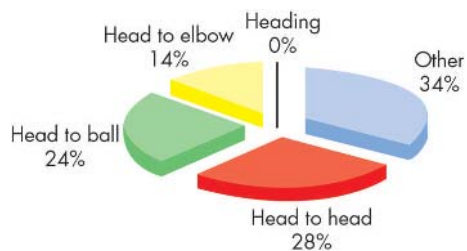
Figure 1 Mechanisms of concussion (from Boden *et al* 1998[13]).

Figure 2 Head injury mechanisms (CPSC, 2002[14]).

indicated that head trauma occurred more frequently from head to head impacts rather than from headings.[10] This is consistent with another study of elite players by Barnes *et al*[15] in which contact from an opposing player was indicated as a source of concussion. Player contact was involved in 32% of cases among the young men and in 71% of cases among the young women. Head injuries resulting from player contact were reported in 79% of cases in a study involving 264 players of all ages and skill levels.[16] The relation between head injuries and player contact is not too surprising given that a large percentage (50–74%) of all injuries in football arise from contact with players.[16–18]

Direct head contact and heading are implicated in acute and chronic head injuries, respectively. However, the injury mechanisms may differ and should be addressed separately in the assessment of protective measures. Head injuries have been noted to stem from one of two categories: head impact with an object (head, elbow, knee, foot, goalpost, ground) and head impact with the ball.[5]

The purpose of this current study was to evaluate the protective capacity of football headgear that has recently entered the market. At the time of this study, three products were offered on the North American market. All claimed to offer head protection while playing football, some specifically stating protection from ball impacts. Regardless of whether that ball impact is based on voluntary heading or inadvertent high speed ball strikes to the head, the primary focus of our current experiments was to investigate the ability of headgear to reduce the injury potential associated with ball to head contact.

However, a high incidence of head to head contact has also been reported by Boden *et al*.[13] A later investigation of game video by Withnall *et al*[19] indicated that nearly a third of all head impacts were due to head to head collisions and another third due to upper extremity impacts. Laboratory simulations of these events revealed that upper extremity impacts posed low concussion risk, but head to head impacts posed high concussion risk. Recent work by Fuller *et al*[20] confirms head to head contact is the most likely scenario for concussion, and upper extremity contact is more likely the cause of head contusion. Therefore, although the focus of our research was ball–head impact, we also investigated the benefit of headgear in head to head impact.

METHODS

Ball to head testing included low speed headers to high velocity ball impacts. Low speed headers may be considered voluntary and high speed ball impacts may be considered inadvertent, where the player is caught unaware. We tested head to head impact in two different orientations and at multiple speeds. For the ball to head impacts, tests were conducted using a human volunteer (up to 8.4 m/s) as well as a surrogate test headform (10–30 m/s). For the head to head tests (2–5 m/s), two biofidelic dummy headforms were used rather than volunteers due to the greater risk of injury from these impacts.
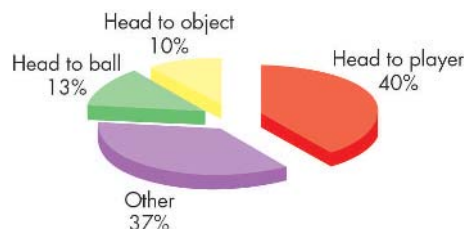
Test samples

The products tested in this study were those current available in North America, and as offered for sale in October 2003. Ten samples of size medium headgear were obtained from Head Blast (St Louis, MO), Full90 performance headgear (San Diego, CA), and Kangaroo Soccer Headgear (Houston, TX). Other models referenced in literature by Broglio *et al*[21] were no longer available at the time of the study. These include the Headers model, which was the former name of the Full90 Sports headgear, and the Protector, which contains an imbedded plastic sheet similar to the Head Blast.

Full90 performance headgear

The Full90 sports headgear (fig 3) provides coverage around the periphery of the head with protective zones on the sides and front of the head. Forehead coverage extends from the brow to the hairline. Ventilation openings are provided in the temporal and rear regions. There are elastic straps at the rear for sizing. Construction of the headgear includes resilient foam which is bonded to a fabric exterior. It is approximately 11 mm thick.

The headgear is advertised to help protect against concussion. Impacts from head to another player, head to ground, and head to goalpost are mentioned as possible sources of impact in football which may lead to concussion. Full90's laboratory tests included head to head, head to goalpost, and head to knee contact. Attenuation was reported to be up to 50% for linear and rotational accelerations.[22] Note that in these tests headgear was fitted to both struck and striking heads. In a research programme sponsored by Full90, the effects of heading were further analysed with a numerical model. The findings stated that the headgear provides protection by reducing the forces to the head.[23] Use of the Full90 headgear was claimed to have no effect on the ball rebound characteristics when tested between 3 m/s and 19 m/s.[24]

Full90 makes no specific claims regarding protection from ball impact. But the product was formerly called "Headers", and patent documentation describes its purpose for heading protection. Based on this, and the notion that consumers

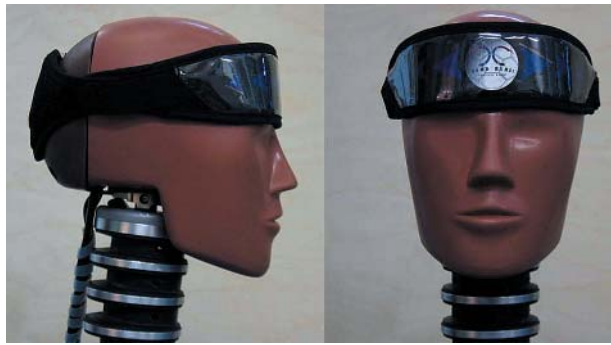**Figure 3** Full90 headgear on Hybrid III.

Figure 4   Head Blast headgear on Hybrid III.

might reasonably presume the product to be intended for heading, we included the Full90 headgear for heading and ball impact tests.

Head Blast

The Head Blast headgear (fig 4) is a wraparound style with protective padding between the brows and hairline and extending laterally to the temple regions. A rear pad is also provided. The front pad is constructed of a resilient foam liner with fabric backing and thin plastic front cover. The primary thickness in this region is approximately 8 mm. A thin plastic sheet is also embedded within the foam pads, presumably to help distribute the load. Hook and loop fasteners are provided on the sides for sizing. Claims for impact protection are not apparent but promotional literature indicates its use primarily for heading.

Kangaroo Soccer Headgear

The Kangaroo Soccer Headgear (fig 5) is similar to that used in martial arts and is primarily intended to provide protection for children and youth players. Coverage extends around the head including around the ears. An elastic chin strap is provided. Construction consists of a resilient foam liner encased in a vinyl dipped protective covering. The overall thickness is approximately 20 mm with the forehead region having additional padding at 28 mm.

The intended use of the headgear is for general player to player impacts as often experienced by less skilled players (Calvin Williams, Kangaroo president, personal communication, 2003). Other promotional information claims to offer protection from ball to head contact as well as other non-specific injuries.

Ball selection

The current study addresses ball impacts for adults. We used a Fédération Internationale de Football Association (FIFA) inspected size 5 ball, Adidas Fevernova Tri-lance, having a mass of 430 g and specified ball pressure of 0.8 bar. Ball pressure was regularly checked during testing.

Test conditions

We conducted all the tests at ambient indoor conditions (19–22 °C, 15–40% relative humidity). All test samples were maintained at these conditions prior to and during testing. An exception to this was where the headgear was worn by the human subject for repeated testing, in which case the headgear remained on his head throughout the test series and may have absorbed some body heat.

High speed video

We recorded the tests with a high speed digital video camera (Motion Scope, Redlake Imaging, USA, model 1000) at a rate

Figure 5   Kangaroo Soccer Headgear on Hybrid III.

of 500 frames per second. The high speed video was used to verify ball trajectory, ball speed, and impact site. The camera was positioned laterally 5.0 m from the volunteer. All analysis was two dimensional in the plane of heading motion. Lateral motion was considered negligible.

Data collection

All data were collected at 10 kHz, following the requirements of SAE J211-1 (Society of Automotive Engineers).[25] All data channels were filtered with CFC1000 anti-aliasing filters. Acceleration data were then digitally post processed to CFC180 for use with the angular acceleration routines.[26]

Test headform and neck

We chose to use a Hybrid III automotive test dummy headform (Denton ATD Inc., Milan, OH) in the evaluation of the headgear because of its human-like response and the availability of literature correlating the head response to injury potential for a wide range of impact conditions. The head anthropometry approximates that of a 50th percentile adult male and has correct mass and mass moment inertial properties for proper dynamics.[27] The biofidelity of the head was based on cadaver head data involving rigid and padded surface impacts with and without skull fracture.[28] Although the biofidelity is primarily based on short duration impacts to the forehead, it has gained wide acceptance for long duration impacts owing to its correct mass properties and use of a biofidelic neck. The kinematics of the neck structure were based on human volunteer data for flexion and extension in the midsagittal plane.[29] The same researchers reviewed the lateral biofidelity in a later study. In all cases, the neck stiffness approximates a tensed state, similar to that experienced when anticipating an impact.
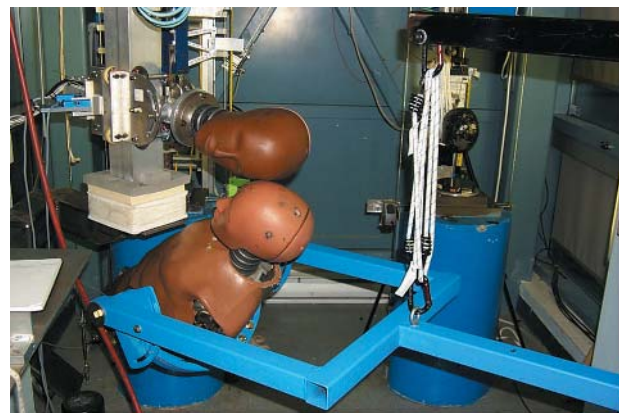

Figure 6   Head to head impact test set-up (front boss to side). Note that the falling head (upper) gets fitted with the headgear, and the lower head/torso may rebound naturally on impact.

Figure 7 Head to head impact test set-up (forehead to rear). The upper head is fitted with the headgear and falls onto the lower head.

For this testing, the headform (DentonATD, model C1846) was instrumented with nine linear accelerometers in the 3-2-2-2 configuration.[30] We used the piezo-resistive Endevco model 7264a-2000 linear accelerometers (Endevco Corp., San Juan Capistrano, CA), which meet the accuracy and frequency response requirements of SAE J211-1. Data processing algorithms implemented in custom Labview routines allow for the complete calculation of linear and angular head accelerations.[30] The custom routines also provided output of various head injury criteria such as the head impact power (HIP).[31]

Head to head testing

Withnall *et al*[19] reported on game video of head impact, supplied by the FIFA Medical Assessment and Research Center (F-MARC), that was reviewed to determine typical impact scenarios. For head to head impact, the most prevalent were the front corner of the striking head to the side of the struck head, and the forehead of the striking head to the rear of the struck head. In that study, head to head game video was analysed and impact speeds up to 2.5 m/s were observed, and head to head laboratory tests performed up to 3 m/s. For purposes of this study, we used the same configurations, but the test speeds were increased up to 5 m/s to investigate the limitations of the headgear. Head to head testing in this manner has been shown to be very repeatable, so multiple repeats were not necessary.[19]

The test was set up such that a Hybrid III head neck system (representing the struck player) was accelerated by gravity alone into contact with a stationary Hybrid III dummy (representing the striking player) that was suspended from an adjustable hoist assembly. Head to head impact on the field occurs typically in a horizontal direction, but since gravity acts vertically, both struck and striking test manne-quins were turned 90°. An illustration of the set-up for the front corner (or front boss) to side is shown in fig 6 and for the forehead to rear in fig 7.

For all tests, only the struck player (that is, falling dummy head) was fitted with protective headgear. Testing was conducted at speeds of 2 m/s, 3 m/s, 4 m/s, and 5 m/s at the two impact sites described. Tests were done first with bare heads, after which each of the three headgears was

Figure 8 Volunteer prepared for heading (bare head configuration shown).

installed. A new headgear was used for each test, such that any potential deterioration of the headgear would not compromise subsequent tests. However, the same headgear was reused for a second impact site.

Four test speeds at two sites, with the bare head plus three headgear, totalled 32 tests.

Volunteer ball heading

For the human heading impacts, ball speed was selected to represent a moderately low level impact experienced during heading (6.4 m/s and 8.2 m/s) and to reduce the potential of discomfort or injury to the volunteer. These speeds were consistent with similar volunteer tests conducted by Shewchenko *et al*.[32] A 30 year old man, having played competitive football in the past and continuing to play recreationally, volunteered to be the test subject for the human ball to head trials. He was 170 cm tall, weighed 77 kg, and had a head circumference of 58 cm. We recruited only one subject for this experiment, on the basis that comparative response between headgear will be similar for any player.

The subject was instrumented with a bite plate on which were mounted two orthogonal accelerometers (Endevco model 7264a-2000) to measure motion in the plane of impact. These accelerometers were not positioned at the centre of gravity of the head, so their data may only be used for comparative purposes between headgear models, not relative to injury potential. Further information is available in the paper by Shewchenko *et al*[32] in this supplement.

The subject was also instrumented with retro-reflective targets to permit measurement of the player kinematics. These were located at the tragion, infra-orbitale and at the location of the accelerometers on the bite plate. Additional targets were affixed to a back bar, which was secured to the subject's back via a series of shoulder and waist straps. A tape switch was affixed to the surface of his forehead to serve as an electronic indicator of ball contact. For tests involving headgear, we taped this switch to the outer surface of the headgear. Two sets of headgear were used, one for each trial speed. For bare head tests, a thin Spandex head cover (used typically for kayaking) was used to hold the switch in place. Figure 8 shows the subject prepared for testing with bite plate installed.

The subject was instructed to head the ball and strike a target which was approximately 50×50 cm, suspended 1 m above the floor, and 4 m away. In this manner there was a reasonable degree of consistency in the trajectory of the ball. We discarded the tests where the target was missed. Five repeats for *both* test speeds were done to average out
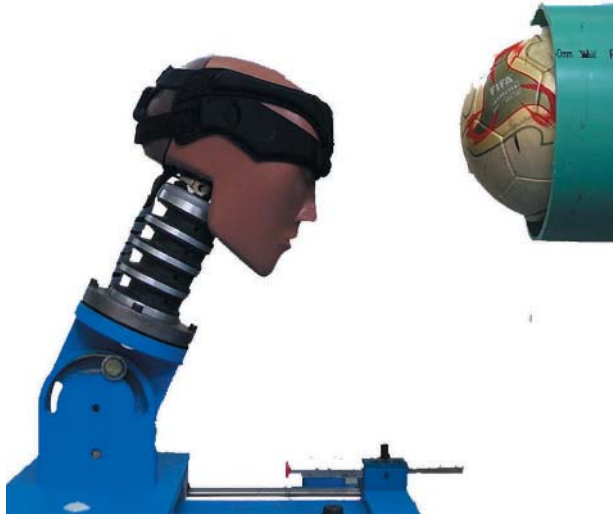
Figure 9 High speed ball cannon set-up relative to test headform (Full90 headgear shown).

Figure 10 Test headform, air cannon and slider system (Kangaroo headgear installed).

30 m/s were done for verification of repeatability. The overall total was 16 high speed tests.

Injury analysis

It is important to note that the headgears used in our study are not crash helmets. They are not intended to mitigate acute brain injury in severe impact conditions. It is necessary to compare headgear performance within the scale, and within the metric, of anticipated injury. Of interest here is the reduction of sub-concussive head impact forces. Newman et al[31] introduced the HIP as a new index relating mild traumatic brain injury to the linear and angular head accelerations and velocities of professional American football players. This function computes the time rate of energy transfer to the head. Logist regression functions were used to approximate the risk of concussion associated with a test score, and the maximum value of power (HIPmax) was found to correlate better with the observed incidence of physician-diagnosed concussion than other commonly referenced parameters such as peak linear acceleration, head injury criteria (HIC), or severity index (SI). A more detailed discussion on the efficacy of HIPmax may be found in the paper by Shewchenko et al[32] in this supplement.

The calculation of HIPmax relies on complete time histories of the centre of gravity linear and angular accelerations and derived velocities.[32] This was not possible with our volunteer heading the ball, since the instrumentation was not at the centre of gravity of his head. Therefore, with volunteer heading data we only used linear acceleration directly from the biteplate to compare the effects of the headgear, whereas we used HIPmax for all tests with the manikin headform to relate impact severity with injury risk.

For the purposes of this study, the 5%, 50%, and 95% risk levels for concussion are considered. The upper and lower 5% of the logist injury risk functions are not used in practice. This range of concussion risk may be interpreted as unlikely, even chances, and almost certain injury. The 5%, 50%, and 95% concussion risk levels associated with these functions are summarised in table 1.

variations from test to test. A powered carriage was used to launch the ball towards the subject in a repeatable manner without ball spin. The launch speed was varied by adjusting how far the carriage was pulled back before release, and the attitude of the launcher could be adjusted to change the ball's trajectory. The launcher was 5.0 m from the subject. This same apparatus has been used in a previous study of heading techniques.[33]

Five repeats at two speeds with a bare head and three headgear totalled 40 subject tests.

High speed ball impact

Typical upper limit ball speeds of 26.8–53.6 m/s have been reported to occur in games whereas the maximum ball speed likely to be headed was less than 18 m/s (65 km/h).[5 11] For the laboratory tests, we chose ball speeds of 10 m/s, 20 m/s, and 30 m/s. The target location was to the forehead of the headform, similar to a heading manoeuvre. Although it is recognised that voluntary heading at 30 m/s is probably atypical, it nevertheless provides a good indication of the headgear's performance trend for accidental impact.

A pressure-venting air cannon was used to propel the ball at these high speeds. We designed a lightweight sabot for a loose sliding fit within this barrel. A hemispherical cavity was machined into the face of the sabot to hold the ball in place with a gentle friction fit. Upon venting of the pressurised air tank, the sabot and ball are accelerated through the barrel, but the sabot is arrested at the mouth of the cannon, allowing the ball to exit at the desired speed. In this manner, exceptional aiming accuracy is achieved without ball spin or air blast. We controlled the exit speed by adjusting and stabilising the pressure in the air tank. The air cannon is shown in fig 9, and it was used for the 20 m/s and 30 m/s tests. The powered carriage was used for the 10 m/s tests.

The same Hybrid III head and neck system from the head to head tests was used. A new set of headgear was used for each of the three trial speeds. The neck was inclined forwards at 25°. The base of the neck was mounted on a linear sliding table (15.8 kg) to allow for rebound of the head neck system on impact. The test set-up is illustrated in figs 9 and 10. All high speed ball strikes were to the forehead of the dummy.

Three ball speeds were tested against the bare head plus three headgear. Two additional runs at both 10 m/s and
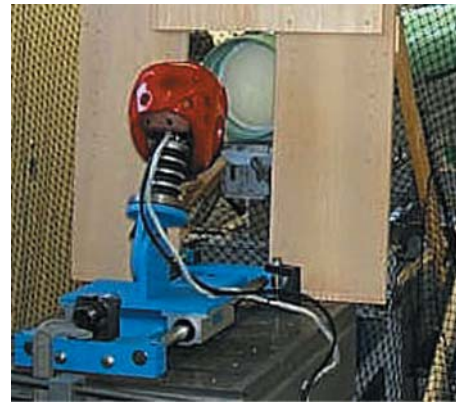
Table 1 Concussion risk for linear and angular accelerations and HIPmax

Risk level	5%	50%	95%
Linear acceleration (g)	40.0	78.0	115.0
HIPmax (kW)	4.5	12.8	21.3

HIPmax, maximum Head Impact Power.

Table 2 Head to head impact test data

Headgear	Speed (m/s)	Front boss to side Peak linear accel. (g)	HIPmax (kW)	HIP concussion risk (%)	Forehead to rear Peak linear accel. (g)	HIPmax (kW)	HIP concussion risk (%)
Bare head	2	31.4	1.5	<5	39.7	2.1	<5
	3	81.4	7.0	10.9	84.1	6.2	8.4
	4	120.2	13.6	55.9	132.3	13.6	55.7
	5	160.9	21.5	>95	168.5	19.8	92.0
A	2	20.6	1.1	<5	23.7	1.2	<5
	3	55.2	4.6	5.0	60.0	4.1	<5
	4	106.2	11.8	40.3	107.2	10.9	32.7
	5	144.4	20.3	93.1	151.1	17.8	85.0
B	2	25.7	1.4	<5	26.3	1.3	<5
	3	59.7	4.9	5.4	68.0	4.8	5.3
	4	97.9	10.4	28.6	116.8	11.8	40.3
	5	140.2	18.0	86.0	157.0	18.6	88.3
C	2	10.1	0.7	<5	13.1	0.7	<5
	3	26.6	2.4	<5	29.2	1.9	<5
	4	69.4	7.8	14.0	61.2	5.8	7.3
	5	88.4	12.4	45.4	92.8	10.8	32.2

accel., acceleration; HIPmax, maximum Head Impact Power.

Statistical analysis

We used Student's t test for determining statistical significance, assuming equal variance and double tail distribution. The t test ($\alpha = 0.05$) was performed using the data analysis package in Microsoft Excel. Equal variance was confirmed by an F test ($\alpha = 0.05$).

RESULTS

To investigate the protective capacity of headgear in general, and not to endorse or criticise particular models, the headgear data have been blinded.

Table 3 Head to head impact test data: percentage difference of headgears relative to the bare head

Headgear	Speed (m/s)	Peak linear accel. % diff.	HIPmax % diff.
Front boss to side			
A	2	−34.4	−21.8
	3	−32.2	−33.9
	4	−11.7	−13.0
	5	−10.3	−5.7
B	2	−18.1	−5.7
	3	−26.7	−30.2
	4	−18.5	−23.7
	5	−12.9	−16.1
C	2	−67.7	−54.7
	3	−67.4	−66.0
	4	−42.3	−42.3
	5	−45.0	−42.2
Forehead to rear			
A	2	−40.2	−44.1
	3	−28.6	−34.6
	4	−19.0	−19.6
	5	−10.3	−10.1
B	2	−33.8	−39.2
	3	−19.1	−22.4
	4	−11.7	−12.9
	5	−6.8	−5.9
C	2	−67.0	−66.5
	3	−65.3	−69.2
	4	−53.8	−57.2
	5	−44.9	−45.2
	Mean	−32.8	−32.6
	SD	20.2	20.2
	CI	8.1	8.1

accel., acceleration; diff., difference; HIPmax, maximum Head Impact Power.

Head to head

Head to head test results for the two configurations at 2–5 m/s are provided in table 2. Included here are maximum linear accelerations, HIPmax, and the risk of concussion related to the maximum power. The bare head impacts for both front boss to side and forehead to rear tests showed unlikely risk of concussion at 2 m/s, approximately 10% risk of concussion at 3 m/s, just over 50% risk at 4 m/s, and virtually certain risk at 5 m/s. The reader is reminded that previous analysis of head to head impacts measured actual game collisions in the region of 2.5 m/s resulting in minor head injuries but no concussions.[19] In all cases, the presence of football headgear reduced the peak acceleration and maximum HIP values, and the corresponding risk of concussion. At 3 m/s, 4 m/s, and 5 m/s concussion risks were reduced to less than 5%, 7.3%, and 32% respectively.

For an overall comparison, the test scores are presented as percentage reduction in table 3. In this table, the percentage values do not relate to concussion risk, merely the change in acceleration or HIPmax attributed to the headgear. Averaging the percentage reduction across all speeds and both configurations provides a general indication of overall benefit, where we make the assumption that these impact speeds and sites would be evenly distributed among the population, and therefore there is no weighting to particular configurations. The presence of headgear in head to head impact reduced peak linear acceleration by −32.8% (95% CI −40.9% to −24.7%) and HIPmax by −32.6% (95% CI −40.7% to −24.5%).

Volunteer heading

For the volunteer heading tests some variability was expected so five repeats were done with the bare head, and then with each of the three headgears. Test results for the 6.4 m/s and 8.2 m/s ball launch speeds are provided in table 4. All data are peak resultant accelerations in the plane of forward heading motion. There was no lateral accelerometer affixed to the bite plate. For these data, accelerations refer to the instrumentation affixed to the bite plate, and do not relate directly to centre of gravity head acceleration. However, they may be used for relative comparison of headgear response on the basis that higher bite plate accelerations indicate higher centre of gravity accelerations and vice versa. A study of American football helmet performance found good correlation between intraoral measurements and cranial response.[34]

Table 4 Volunteer heading biteplate acceleration data*

		Ball speed 6.4 m/s	8.2 m/s
Bare head	Mean	5.4 g	7.6 g
	SD	1.5 g	1.6 g
A	Mean	6.8 g	5.7 g
	SD	0.6 g	1.8 g
	t stat	−2.013	1.699
	p	0.079	0.128
B	Mean	5.1 g	6.1 g
	SD	1.6 g	1.1 g
	t stat	0.244	1.644
	p	0.813	0.139
C	Mean	5.5 g	6.8 g
	SD	1.1 g	1.2 g
	t stat	−0.154	0.862
	p	0.881	0.414

*n = 5 for each mean, comparing headgears to bare head (two tailed t test, df = 8, tcritical = 2.306).

Table 5 High speed ball impact test results

Headgear	Ball speed (m/s)	Peak linear accel. (g)	HIPmax (kW)	HIP concussion risk (%)
Bare head	10	16.4	0.6	
	10	15.1	0.6	
	10	15.5	0.5	
	Mean	15.7	0.5	<5
	SD	0.63	0.02	
	95% CI ±	0.7	0.02	
	20	32.9	1.9	<5
	30	57.5	5.0	
	30	52.8	4.5	
	30	55.9	4.7	
	Mean	55.4	4.7	5.3
	SD	2.4	0.29	
	95% CI ±	2.7	0.32	
A	10	15.5	0.5	<5
	20	29.4	1.7	<5
	30	59.9	5.3	6.1
B	10	14.5	0.5	<5
	20	32.9	1.9	<5
	30	55.1	4.3	<5
C	10	15.2	0.5	<5
	20	31.2	2.0	<5
	30	58.7	5.9	7.7

accel., acceleration; HIPmax, maximum Head Impact Power.

For each headgear and ball speed, Student's *t* test was performed for the five repeats, assuming equal variance (verified by *F* test) and two tailed distribution (df = 8, *t*critical = 2.306). For both the 6.4 m/s and 8.2 m/s ball speeds no significant differences were observed between the bare head and the various headgears (p>0.05). This suggests that the headgear provided no significant reduction in head impact from ball heading.

High speed ball impact

The results of the high speed ball impact tests are summarised in table 5. Bare head tests were repeated three times at the 10 m/s and 30 m/s speeds to check the repeatability of the experimental set up, which was very good and within the expected margin for this type of testing. For all further comparisons, single tests were run, and are compared with the average result of the above three 10 m/s and 30 m/s repeats. Two immediate observations are: firstly, the acceleration and HIPmax levels are generally very low, and secondly, that the risk of concussion is very low. Even at the 30 m/s ball speed, which is conceivably a speed higher than voluntary heading, concussion risk was below 6% for the bare head and below 8% for the various headgear. In

Table 6 High speed ball impact percent difference for headgears relative to bare head

Headgear	Ball speed (m/s)	Peak linear accel. % diff.	HIPmax % diff.
A	10	−1.3	−12.4
	20	−10.7	−8.7
	30	8.0	11.6
B	10	−7.3	−15.6
	20	0.1	0.3
	30	−0.5	−9.9
C	10	−3.4	−13.5
	20	−5.3	7.3
	30	5.9	25.9
	Mean	−1.6	−1.7
	SD	6.0	14.1
	95% CI±	3.9%	9.2%

accel., acceleration; diff., difference; HIPmax, maximum Head Impact Power.

Figure 11 Snapshot from high speed video of 30 m/s ball impact (Full90 headgear).

some cases the headgear reduced scores, and in other cases they were actually higher. This is more obvious when the data are converted to percentage difference for the headgear compared with the bare head (table 6). Combining the headgear percentage differences showed no benefit in the 95% confidence intervals: neither for peak acceleration (−5.5% to 2.3%) nor for HIPmax (−10.9% to 7.5%). Note that taking an overall average of percentage differences assumes that the headgear models and ball speeds are equally distributed among the population, such that no configuration is weighted in overall benefit.

DISCUSSION

None of the three headgears tested appears to be particularly effective at reducing the impact from ball contact, whether it is voluntary or incidental. Although only one volunteer represented the human response, tests with the Hybrid III headform showed similar results. Test data revealed no observable benefit in head acceleration or maximum HIP from the wearing of football headgear. In fact in some cases, especially the 30 m/s ball impact, the results were slightly worse wearing the headgear than without. But these differences were small and trivial. On the contrary, all of the headgears afforded protection against direct head to head impact. Overall reductions across all headgear were approximately 33% for both linear acceleration and HIPmax.

So why are these headgear not effective in heading? The answer lies the relative stiffness of the objects involved in the ball head collision. In simple terms, the human head is stiffer than the ball. Therefore on impact, the ball deforms more than the head. A dramatic example of this is shown in fig 11: a video frame of a 30 m/s ball to forehead impact, wearing a headgear. Digitisation of the high speed video revealed approximately 100 mm of ball compression onto the headform. Traditional crash helmets or contact sport helmets typically include a layer of crushable material that is softer than both the skull and the object being struck (for example, asphalt) such that it becomes the sacrificial weak layer. This protective layer deforms, causing energy to be dissipated and force on the head reduced. Reduced force means reduced acceleration and reduced injury risk. However, in this ball impact, the amount of ball deformation is nearly ten times greater than the nominal thickness of the headband. Whether the headband material was very stiff (and did not crush at all) or very soft (and flattened completely), the overall performance difference would be minimal.

If one were to imagine that same test, but with a stiffer ball, which hence deformed less on impact, the thickness of

What is already known on this topic

Football head protectors have recently entered the market-place. Media attention has focused on potential injury from ball heading and other situations of head impact in football. Opponents feel that headgear are not warranted and will have a negatively effect on the game. Debate continues on whether or not the time has come for protective headgear in football.

What this study adds

If headgear is to be worn in football, the effectiveness of that headgear and its limitations must be understood. Three commercially available football headgear were tested. They were found to offer no protection from ball heading and high speed ball impact, but did offer some protection against impact from another player's head.

the padding would be a larger fraction of the total deformation (that is, padding + ball). Make the ball even stiffer, and the contribution of the padding would continue to be more effective. Now imagine the ball being as stiff as another head, similar to the head to head test. In this situation, the headband is the only item deforming and its presence makes a positive difference compared with no headband at all, as seen in the laboratory tests. A headband is effective when inserted between two stiff objects, but is ineffective when one of the objects deforms more than the headband. A headband would need to be thicker, and likely softer, to reduce successfully the heading induced acceleration. This would almost certainly lead to negative acceptance and negative effect on heading control and power. Considering the much higher potential for concussion from head to head contact than ball contact, and given that all headgears demonstrated the ability to mitigate head hits, there exists a positive potential for headgear to be redesigned specifically to protect against head to head or other non-ball related impact. To this end, areas of coverage might be tailored to those regions most vulnerable to non-ball contact, and padding material and stiffness optimised accordingly.

CONCLUSION

In the current test programme, we measured the performance of headgear against ball impact conditions of low speed volunteer heading (6.4–8.2 m/s), high speed ball dummy head impact (10–30 m/s) and against direct dummy head to head contact (2–5 m/s).

The findings of the low and high speed ball impacts, with a human subject and dummy headform, respectively, showed that all headgears did not reduce the impact response of the head. The head acceleration responses were within the variability noted for the unprotected condition. High speed video confirmed that the ball undergoes much larger deformation than the headband. The headgears tested are not effective because it is the ball which dominates the impact response. The head responses observed in all ball to head tests indicated a low risk of concussion even at the highest speed tested.

The findings of the head to head impacts show that headgear provides a measurable improvement in head response for the two impact sites tested. Due to the relatively high stiffness of the colliding heads, the introduction of a compliant headgear helps attenuate the impact by dissipation of energy. A prime original intent of football headgear was for protection while heading, however, it is reasonable that further performance gains might be achieved if materials are optimised for impact with the head or other rigid objects. Further effort shall be required to define other protective aspects of football headgear and develop relevant performance specifications and product standards.

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COMMENTARY

This article is timely and will answer some of the questions surrounding the effectiveness of football (soccer) headgear in preventing concussions. For clinicians who provide care for athletes, this article will have an immediate impact on our practice, as it provides valuable information which will help us to answer questions and counsel patients. This is the first published research testing the effectiveness of available football headgear to prevent concussions in head to head contacts, as well as head to ball impacts. The results provide clinicians and researchers a quantifiable measure of headgear effectiveness in decreasing the risk of concussion in the laboratory. The research also reveals what researchers have previously shown in epidemiological data: that a ball to head contact is usually not of sufficient force to cause a concussion, either with or without a headgear in place. Where football players need most protection from concussions is during head to head contact with other players. These headgear seem to provide protection during these high risk collisions as tested in the laboratory setting.

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