Evaluation of an In-Ear Sensor for Quantifying Head Impacts in Youth Soccer

Stian Bahr Sandmo,*†‡ MD, Andrew S. McIntosh,§||¶ PhD, Thor Einar Andersen,† MD, PhD, Inga Katharina Koerte,#** MD, and Roald Bahr,† MD, PhD
Investigation performed at the Oslo Sports Trauma Research Center, Department of Sports Medicine, Norwegian School of Sport Sciences, Oslo, Norway

Background: Wearable sensor systems have the potential to quantify head kinematic responses of head impacts in soccer. However, on-field use of sensors (eg, accelerometers) remains challenging, owing to poor coupling to the head and difficulties discriminating low-severity direct head impacts from inertial loading of the head from human movements, such as jumping and landing.

Purpose: To test the validity of an in-ear sensor for quantifying head impacts in youth soccer.

Study Design: Descriptive laboratory study.

Methods: First, the sensor was mounted to a Hybrid III headform and impacted with a linear impactor or a soccer ball. Peak linear acceleration (PLA), peak rotational acceleration (PRA), and peak rotational velocity (PRV) were obtained from both systems; random and systematic errors were calculated with Hybrid III as reference. Then, 6 youth soccer players wore sensors and performed a structured training protocol, including heading and nonheading exercises; they also completed 2 regular soccer sessions. For each accelerative event recorded, PLA, PRA, and PRV outputs were compared with video recordings. Receiver operating characteristic curves were used to determine the sensor's discriminatory capacity in both on-field settings, establishing cutoff values for predicting outcomes.

Results: For the laboratory tests, the random error was 11% for PLA, 20% for PRA, and 5% for PRV; the systematic error was 11%, 19%, and 5%, respectively. For the structured training protocol, heading events resulted in higher absolute values (PLA = 15.6g ± 11.8g) than nonheading events (PLA = 4.6g ± 1.2g); the area under the curve was 0.98 for PLA. In regular training sessions, the area under the curve was >0.99 for PLA. A 9g cutoff value yielded a positive predictive value of 100% in the structured training protocol versus 65% in the regular soccer sessions.

Conclusion: The in-ear sensor displayed considerable random error and substantially overestimated head impact exposure. Despite the sensor's excellent on-field accuracy for discriminating headings from other accelerative events in youth soccer, absolute values must be interpreted with caution, and there is a need for secondary means of verification (eg, video analysis) in real-life settings.

Clinical Relevance: Wearable sensor systems can potentially provide valuable insights into head impact exposures in contact sports, but their limitations require careful consideration.

Keywords: TBI; repetitive; soccer; subconcussive; wearable; accelerometer

Repetitive head impacts in the "subconcussive" range (ie, head impacts without immediate symptoms) are common in soccer, where purposeful and unprotected heading of the ball is an integral part of the game. There is evidence of long-term structural and functional brain alterations among soccer players.[4-6,9] Moreover, recent studies suggested a potential effect on cognitive function among children and adolescents during a vulnerable period of brain maturation.[7,22] However, the link between exposure to repetitive head impacts and brain alterations is still controversial and remains to be elucidated. In this context, accurate measurement of head impact exposure in soccer is a key challenge when investigating the effect of head impact exposure on brain health.

Wearable sensor systems, such as accelerometers/gyroscopes, can potentially provide valuable insights into the dynamics of single and repetitive head impacts. However, several issues have made quantifying head impact exposure challenging, despite the multiple systems currently available.[12,13] A central issue has been poor sensor

The American Journal of Sports Medicine
1–8
DOI: 10.1177/0363546519826953

coupling with the head; methods such as skin patches and skull caps are subject to relative motion between the device and the skull and are therefore not able to accurately measure head impact exposure in vivo.[20] This issue extends beyond erroneous outputs for direct head impacts: failure to discriminate these from non–head impact accelerative events typically seen during game play (running, jumping, tackling, etc) also leads to high false-positive rates.[2,14] Thus, previous studies concluded that secondary means of verification, such as video analysis, are needed to verify whether the events recorded actually represent head impacts.[2,8,14] This makes surveillance of exposure in large-scale cohort studies considerably more demanding.

In-ear sensor systems have recently become commercially available, aiming to improve skull coupling by custom-molded placement in the bony portion of the external ear canal. However, before usage in prospective cohort studies, it is necessary to evaluate their performance in both a laboratory setting and an on-field setting. The aim of this study was to test the validity of a new in-ear sensor for quantifying head impacts in youth soccer.

METHODS

Study Design and Participants

This study was conducted in 3 phases: (1) validation of the in-ear sensor in a controlled laboratory setting, (2) controlled on-field evaluation of its ability to differentiate headings from other accelerative events typically seen in soccer, and (3) on-field evaluation in regular soccer training sessions. In phases 2 and 3, 6 male youth soccer players participated (mean ± SD age, 15.3 ± 0.3 years; height, 170.3 ± 5.0 cm; mass, 54.8 ± 6.1 kg), with all playing at the regional elite youth level in Norway during the 2017 season. The ethics committee at the Norwegian School of Sport Sciences approved the study, and written informed consent was obtained from the participants and their legal guardians.

In-Ear Sensor

MV1 (MVTrak) is a sensor system designed for custom-molded placement in the left external ear canal to optimize coupling to the head. A small lumen runs through the sensor to allow air conduction, limiting hearing loss to approximately 3 dB. The sensor samples linear acceleration and rotational velocity data at 1000 Hz, filtering the data with a phaseless 300-Hz 8-pole low-pass Butterworth filter to remove noise; rotational acceleration is calculated by differentiating these filtered rotational velocity data. The sensor then provides a time-stamped output of peak linear acceleration (PLA), peak rotational velocity (PRV), and peak rotational acceleration (PRA) for all accelerative events exceeding 3*g* (ie, nominal head impacts), followed by a 250-millisecond latency period before another impact can be registered. The sensor stores event-specific data on a microchip and connects via USB to a computer for download. Raw data are uploaded to the MVTrak server before being processed by the producer's algorithm. These data can then be downloaded for each player (ie, sensor) as time-stamped and event-specific summaries in Excel charts, including PLA, PRV, PRA, and the individual kinematic components of each accelerative event.

Experimental Procedures

Phase 1: Laboratory Validation. The MV1 sensor was mounted at the ear region of an in-calibration Hybrid III (HIII) head and neck assembly (Humanetics). Three mounting configurations were assessed: (1) a custom-made flat MV1 sensor (MV1 flat) attached with double-sided tape and reinforced with external taping to minimize relative motion between the HIII headform and the sensor and to optimize the coupling conditions for assessment as an alternative to in-ear mounting; (2) a regular in-ear MV1 (MV1 in-ear) firmly placed in a tight canal on the HIII headform, representing an artificial ear canal; and (3) a regular in-ear MV1 (MV1 loose) loosely placed by expanding the same canal (Figure 1). We created the canal by carving out a piece of the artificial skin covering the HIII headform. The tight canal's diameter was slightly smaller than the sensor's, only enough to allow the compliant properties of the rubber to expand and create a snug fit, mimicking real-life custom-molded placement; the expanded canal's diameter was slightly larger (2-3 mm) than the sensor's, allowing slight relative motion for the sensor. The HIII head was instrumented at its center of

*Address correspondence to Stian Bahr Sandmo, MD, Oslo Sports Trauma Research Center, Department of Sports Medicine, Norwegian School of Sport Sciences, PB 4014 Ullevål Stadion, NO-0806 Oslo, Norway (email: s.b.sandmo@nih.no).
†Oslo Sports Trauma Research Center, Department of Sports Medicine, Norwegian School of Sport Sciences, Oslo, Norway.
‡Faculty of Medicine, University of Oslo, Oslo, Norway.
§School of Engineering and ACRISP, Edith Cowan University, Joondalup, Australia.
||Monash University Accident Research Centre, Monash University, Melbourne, Australia.
¶McIntosh Consultancy and Research, Sydney, Australia.
#Department of Child and Adolescent Psychiatry, Psychosomatic, and Psychotherapy, Ludwig-Maximilian-University, Munich, Germany.
**Department of Psychiatry, Psychiatry Neuroimaging Laboratory, Brigham and Women's Hospital, Harvard Medical School, Boston, Massachusetts, USA.

One or more of the authors has declared the following potential conflict of interest or source of funding: This work was part of the study RepImpact and was funded by ERA-NET NEURON, the German Federal Ministry of Education and Research, and the Research Council of Norway. The Oslo Sports Trauma Research Center has been established at the Norwegian School of Sport Sciences through generous grants from the Royal Norwegian Ministry of Culture, the South-Eastern Norway Regional Health Authority, the International Olympic Committee, the Norwegian Olympic Committee and Confederation of Sport, and Norsk Tipping AS. Phase 1 laboratory tests were conducted at the RMS Crashlab, Sydney, Australia. AOSSM checks author disclosures against the Open Payments Database (OPD). AOSSM has not conducted an independent investigation on the OPD and disclaims any liability or responsibility relating thereto.

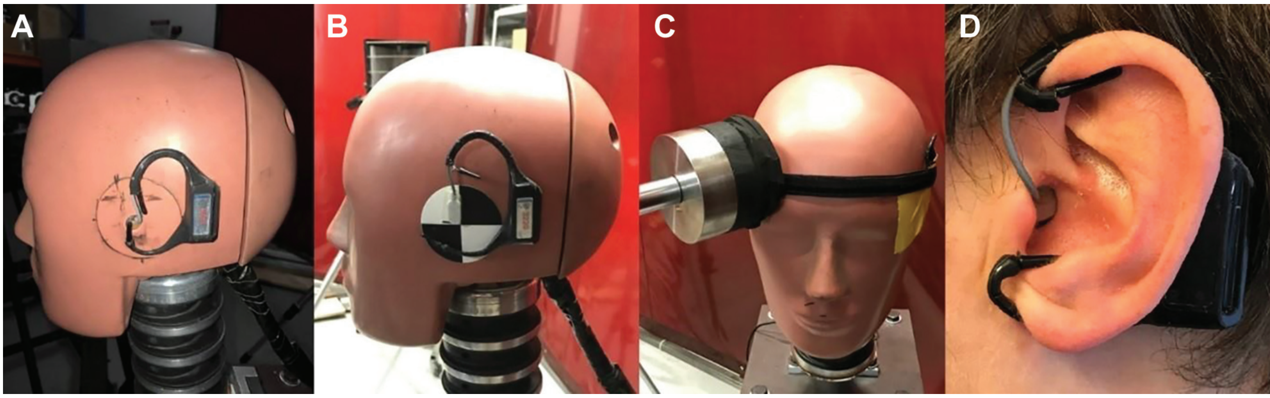

Figure 1. (A) Mounting of the MV1 in-ear and (B) MV1 flat on the Hybrid III headform. (C) A sample setup for right frontal impacts with the padded impactor striking from a 45° angle. (D) MV1 in a real-life setup.

mass with an in-calibration triaxial linear accelerometer (Endevco; Meggitt Sensing Systems) and triaxial angular velocity sensor array (Diversified Technical Systems), sampling at a rate of 20 kHz. Linear acceleration and angular velocity data were filtered with a SAE CC1000 filter and a SAE CC180 filter[15] (DIAdem 2011; National Instruments), respectively, before computing a preliminary set of PLA and PRV values for each impact. PRA values were calculated by differentiating the filtered angular velocity data. HIII-measured impact characteristics were considered reference values for MV1 flat; for evaluating between-sensor agreement, MV1 flat was considered reference for MV1 in-ear. The HIII was impacted at selected locations over a range of magnitudes with a linear impactor with a stiff interface, a linear impactor with a compliant interface, or a FIFA-approved soccer ball inflated to 11 psi (Table 1). Each test was videoed with a GoPro HERO5 Black camera, recording at 240 Hz.

Phase 2: Controlled On-Field Evaluation. The 6 participants were invited to complete a structured training protocol in a controlled setting twice while wearing a custom-molded MV1 in the left ear canal. The protocol was designed and supervised by research staff with long-standing experience in soccer (S.B.S. and T.E.A.) and consisted of 5 heading and 6 nonheading exercise drills typical for soccer. Heading exercises included finishing headers, redirectional headers, long direct headers, short direct headers, and headers from in-air duels. Nonheading exercises included shoulder-to-shoulder collisions, forceful shooting, redirectional running with maximal intensity, short straight sprinting with maximal intensity, falling abruptly forward on the ground and landing on out-stretched arms, and in-air duels without ball contact (losing the duel).

Phase 3: In-Training On-Field Evaluation. The participants wore the sensors for 2 regular training sessions with their team. The sessions were instructed by their regular coaching staff and included warm-up, passing and playing drills, and regular play in teams.

Phases 2 and 3 were performed on artificial turf in an outdoor setting. Video recordings were obtained with 2 digital video cameras (1080p, 50 fps), placed to capture all movements on the pitch to subsequently verify and classify events.

Data Analyses

For the laboratory validation, the HIII kinematic time histories (eg, linear acceleration) were reviewed for comparison with the high-speed video of each test. The aim was to review the preliminary PLA, PRA, and PRV values for each test and to identify the peak values directly related to the initial interaction between the impactor/soccer ball and the HIII headform. After review by S.B.S. and A.S.M., a final set of HIII PLA, PRV, and PRA values was determined.

To estimate the accuracy of the MV1 sensor for different impact types, locations, and mounting configurations, we calculated its random and systematic error. The random error was calculated by first dividing the SD of the mean difference between the MV1 and the reference (HIII) by the square root of the number of measurements ($n = 2$); this value was then divided by the mean of the combined measurements, expressing the random error as a percentage.[16] The systematic error was calculated as the mean difference between the sensor and the reference, divided by the mean reference value. Expressed as percentages, positive and negative results indicate systematic overestimation and underestimation by the MV1, respectively. For the soccer ball impacts, MV1 flat and MV1 in-ear were mounted to the HIII simultaneously; agreement between the sensors was expressed with the same formulas, with MV1 flat as the reference.

For the structured training protocol (phase 2), the individual events of each exercise drill were used as reference and compared with the time-stamped outputs from the sensors. If an event failed to exceed the sensor's $3g$ threshold and was therefore not recorded, kinematic values were set as follows to be included in later analyses: PLA = $3.0g$, PRV = 3.0 rad/s, and PRA = 200 rad/s^2. These values were set arbitrarily, under the assumption that these events involved slightly lower values than the lowest-magnitude events recorded from the sensor; this was done to include them in the receiver operating characteristic (ROC) analyses. For the regular training sessions (phase 3), all head impacts were first identified on video to be included in the analyses; they were then compared with their potential

TABLE 1
Hybrid III Headform (Reference) vs MV1 Flat With Random and Systematic Errors of PLA, PRA, and PRV Values by Impact Type and Location[a]

Impact Type: Location	Impacts, n	Range			PLA, %		PRA, %		PRV, %	
		PLA, *g*	PRA, rad/s^2	PRV, rad/s	Random Error	Systematic Error	Random Error	Systematic Error	Random Error	Systematic Error
Linear impactor										
Frontal	37	26-132	1121-6901	12-23	3	4	14	13	1	1
Right frontal	21	27-110	1755-8030	12-20	9	28	18	21	6	9
Right zygomatic	12	27-138	1835-5087	16-26	5	6	11	45	1	5
Right temple	13	25-144	1668-11,537	11-20	12	–4	13	–6	7	8
Total	83	25-144	1121-11,537	11-26	10	8	18	15	5	4
Soccer ball										
Frontal	9	9-20	997-2203	5-11	17	33	30	54	2	–1
Right frontal	7	13-22	958-4638	7-13	16	67	29	40	11	16
Frontal/crown	10[b]	13-26	1362-3343	7-14	17	39	38	39	10	6
Face	3	11-19	722-3352	6-10	15	40	18	6	10	11
Total	29[b]	9-26	722-4638	5-14	18	45	33	39	10	7

[a]PLA, peak linear acceleration; PRA, peak rotational acceleration; PRV, peak rotational velocity.
[b]PRA and PRV values were excluded for 1 impact.

time-stamped sensor outputs. All other nominal head impact events recorded by the sensors (ie, either non–head impact accelerative events or spurious events) were then classified according to the video.

SPSS (v 24; IBM) was used for all statistical analyses.

RESULTS

Phase 1: Laboratory Validation

For MV1 flat, 112 impacts were included for final analyses (Table 1). When reviewing HIII outputs, we excluded angular kinematic data (ie, PRA and PRV) from 1 of the 112 impacts, since we were unable to identify the appropriate initial peak values. Furthermore, for 1 series of consecutive impacts (n = 12)—all within the same period with identical setup on the same afternoon—angular kinematic values (PRA and PRV) from MV1 flat were recognized as severe outliers (with values ranging from 4 to 13 times higher than the reference). Upon our request, the MV1 producer reviewed the data for these specific impacts and suspected that vibrations between the MV1 flat and the HIII were the cause. We replaced these data points with outputs from MV1 in-ear from the same impacts.

As shown in Figure 2, PLA values showed the strongest correlation, followed by PRV and PRA. The random error for all impacts was 11% for PLA, 20% for PRA, and 5% for PRV. The systematic error was 11% for PLA, 19% for PRA, and 5% for PRV. The random error varied with impact type and location, consistently overestimating PLA, PRA, and PRV values (Table 1). When agreement was tested between MV1 flat and MV1 in-ear for the soccer ball impacts (n = 29 for PLA, n = 28 for PRA and PRV), with MV1 flat as reference, the random error was 6% for PLA, 20% for PRA, and 6% for PRV; the systematic error was –5% for PLA, –23% for PRA, and –3% for PRV.

For MV1 loose, we replicated 7 right frontal impacts and 1 frontal impact (HIII PLA range, 29*g*-122*g*) also used for mounting configuration 2 (ie, MV1 in-ear). As compared with MV1 in-ear, the loose coupling in mounting configuration 3 led to an increase in the random error from 10% to 14% for PLA, 10% to 55% for PRA, and 7% to 20% for PRV. Systematic error increased from 17% to 33% for PLA, 19% to 202% for PRA, and 13% to 32% for PRV.

Phase 2: Controlled On-Field Evaluation

All 6 participants completed each exercise drill at least once, with the number of events obtained per drill ranging from 44 to 180. Heading events (n = 431) resulted in higher average values for all 3 variables (PLA = 15.6*g* ± 11.8*g*, $P < .001$; PRA = 10,543 ± 10,854 rad/s^2, $P < .001$; PRV = 35.1 ± 18.3 rad/s, $P < .001$) as compared with nonheading events (n = 750, PLA = 4.6*g* ± 1.2*g*, PRA = 1095 ± 823 rad/s^2, PRV = 9.8 ± 4.6 rad/s). ROC curve analyses revealed an area under the curve (AUC) of 0.98 (95% CI, 0.98-0.99; $P < .001$) for PLA, 0.99 (95% CI, 0.99-1.00; $P < .001$) for PRA, and 0.97 (95% CI, 0.96-0.98; $P < .001$) for PRV. Figure 3 shows the distribution of peak values for each specific exercise.

Phase 3: In-Training On-Field Evaluation

Five participants completed one or both of the regular training sessions, and from the resulting 8 sessions, the MV1 sensors recorded 2039 nominal head impact events. Of these, 15 events were confirmed on video analysis to be direct head impacts (PLA = 20.7*g* ± 10.6*g*, $P < .001$; PRA = 14,541 ± 7994 rad/s^2, $P < .001$; PRV = 43.5 ± 16.4 rad/s, $P < .001$), all of them attributed to purposeful heading of the ball. No other head impacts were identified on video. The remaining 2024 events were triggered by

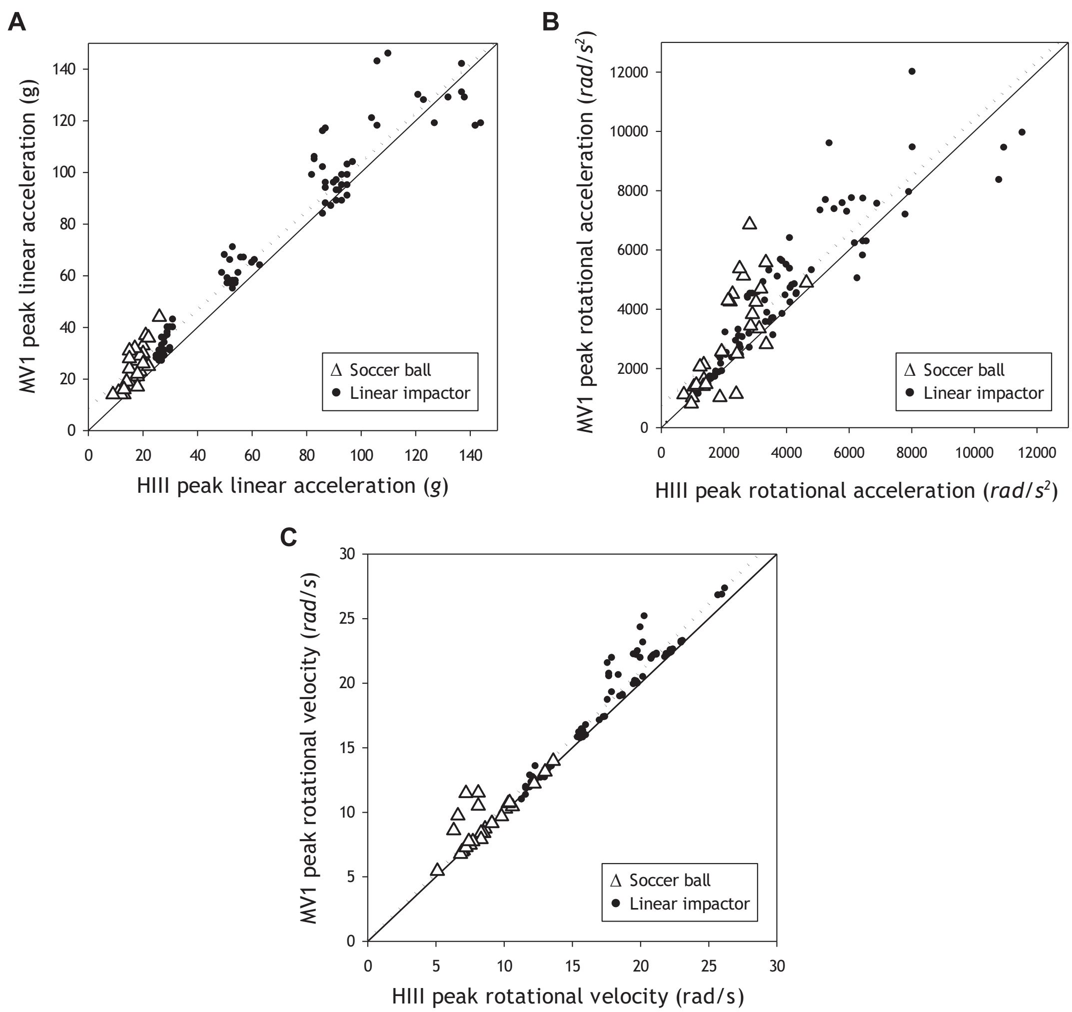

Figure 2. (A) Peak linear acceleration, (B) peak rotational acceleration, and (C) peak rotational velocity from MV1 flat, plotted against the reference (Hybrid III headform). Linear regression lines (dotted) with reference lines (solid) are for all head impacts combined (ie, with linear impactor and soccer ball).

non–head impact events, such as jumping, tackling, running with change of direction, and touching or losing the sensor (PLA = 4.0g ± 3.1g, PRA = 835 ± 2541 rad/s^2, PRV = 7.4 ± 4.9 rad/s), resulting in an AUC >0.99 (95% CI, 0.99-1.00; P < .001) for PLA, PRA, and PRV. Tables 2 and 3 show the sensitivity and positive predictive value of different cutoff values for PLA and PRV for the structured training protocol and the regular training sessions.

DISCUSSION

In this study, we found that the in-ear sensor systematically overestimated head kinematic parameters and with a considerable random error (phase 1). Still, the accuracy for discriminating headers from non–head impact accelerative events in a controlled on-field setting was excellent (phase 2). However, as the proportion of head impacts (ie, headers) was relatively low as compared with non–head impact events, false-positive results nevertheless remained a challenge in the real-life setting (phase 3).

Obtaining accurate results from compact wearable sensor systems is difficult, as shown by Cummiskey et al[3] and others.[11,18,19] In a recent review, Patton[13] described multiple examples of large discrepancies even in controlled laboratory settings. In the laboratory validation (phase 1), we therefore aimed to test the technical performance of the in-ear sensor, optimizing all factors, including coupling to the head. We found a consistent systematic overestimation for all peak values (PLA, PRA, and PRV) and with a considerable random error, varying with impact type and location. Even though the exact reasons are uncertain, several previously recognized technical limitations, such as low sampling rate (1 kHz for the in-ear sensor vs 20 kHz for the reference system), might account for some of the discrepancy. The observation that the PRA component generally performed poorer than PLA and PRV simply reflects that it is derived from PRV, rendering it more susceptible to

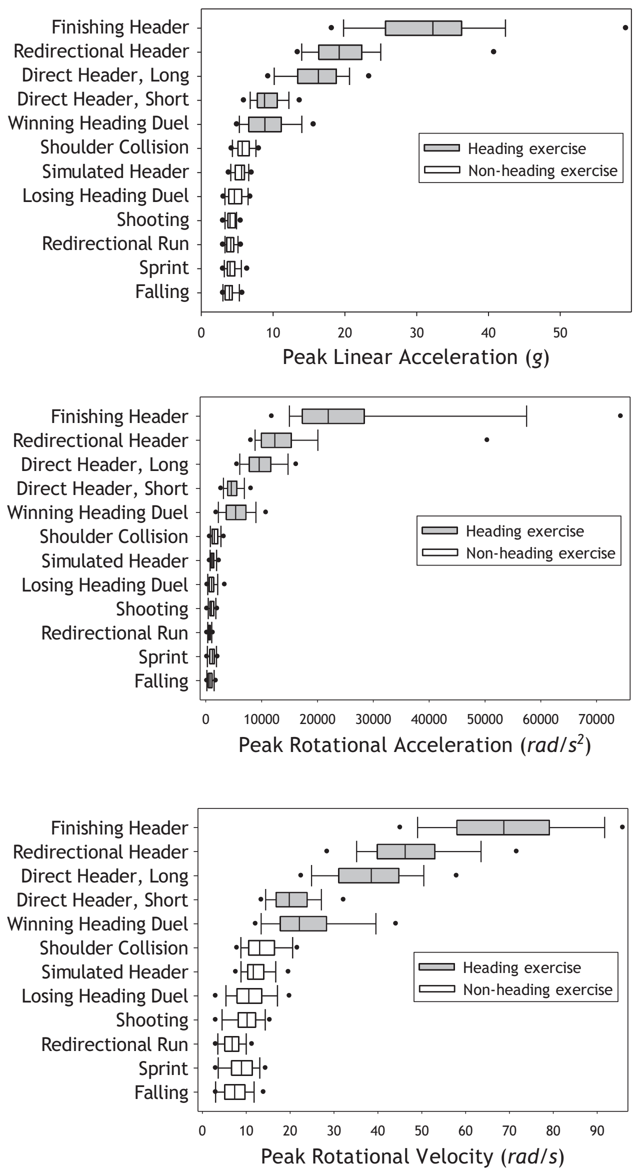

Figure 3. Box plots showing median value and interquartile range of peak linear acceleration, peak rotational acceleration, and peak rotational velocity from MV1 for the exercises from the structured training protocol. The left and right black dots represent the 5th and 95th percentiles, respectively.

noise and to the relatively low sample rate. This is consistent with the finding that PRA values also displayed considerably poorer agreement between sensors (approximately 80%) as compared with both PLA and PRV (approximately 95%). As an additional barrier, algorithms of any externally mounted system need to correct for its relative position on the head to measure what is happening at the center of mass.

As we were interested in how on-field conditions could affect sensor performance in phases 2 and 3, we included a loose mounting configuration in phase 1. The idea was to test how poor coupling could affect the inherent issues

TABLE 2
MV1 Sensitivity and Positive Predictive Value for Classifying Accelerative Events as Head Impacts (Headers) or Non–head Impacts for Different Peak Linear Acceleration Cutoff Values

	Sensitivity, %		Positive Predictive Value, %	
Cutoff Value, *g*	Training Protocol	Regular Training	Training Protocol	Regular Training
>6	96	100	82	22
>7	90	93	93	37
>8	83	87	98	50
>9	73	87	100	65
>10	65	87	100	68

TABLE 3
MV1 Sensitivity and Positive Predictive Value for Classifying Events as Head Impacts (Headers) or Non–head Impacts for Different Peak Rotational Velocity Cutoff Values

	Sensitivity, %		Positive Predictive Value, %	
Cutoff Value, rad/s	Training Protocol	Regular Training	Exercise Protocol	Regular Training
>10	99	100	57	4
>15	92	100	82	18
>20	75	93	94	52
>25	61	93	100	78
>30	52	80	100	75

described here. With an unfavorable effect on both systematic and random errors for all variables, we observed a 10-fold increase in the systematic error for PRA. We believe that this effectively illustrates why one should interpret absolute kinematic values from sensor systems in contact sports with caution. We suspect that the main explanation for some of the very high on-field values observed (Figure 3) is a combination of inherent systematic overestimation and poor head coupling. Arguably, a mean value of well over 20,000 rad/s^2 for finishing headers almost certainly represents a gross overestimation, based on previous biomechanical studies from heading in soccer and mild traumatic brain injuries[1,10,17,21]; the players considered the exercise to be in the upper but normal heading severity range.

Press and Rowson[14] recently quantified head impact exposure in collegiate women's soccer using a skin patch placed behind the ear. They observed that the recorded number of head impacts vastly exceeded those confirmed on video, concluding that data from head impact sensors warrant careful interpretation when used in automated settings. Cortes et al[2] drew similar conclusions when measuring head impact exposure in lacrosse—with both studies highlighting the need to classify accelerative events with, for example, video analysis. Thus, the main objective of the structured training protocol (phase 2) was to evaluate the in-ear sensor's capacity to discriminate head impacts from non–head impact accelerative events. By classifying all recorded accelerative events into these 2 main categories—in both the structured training protocol and the regular training sessions—our results showed that the sensor displayed an excellent discriminatory capacity. However, despite the ability to maintain high sensitivity and specificity, there is a crucial difference between the on-field settings, with real-life implications. In the structured training protocol, it was possible to use a cutoff value (eg, 9*g*) (see Table 2) yielding 100% positive predictive value while maintaining sensitivity >70%. Although such a scenario misses many head impacts in the lowest range, one can safely conclude that any event above this threshold is actually caused by a direct impact to the head, obviating secondary means of verification (eg, video). We were unable to replicate this finding in the regular training sessions (phase 3) owing to spurious non–head impact events, such as touching the sensor or dropping it on the ground, recording values as high as 65*g* and 124*g*. Tables 2 and 3 illustrate the difficulties of identifying a PLA or PRV cutoff value in a real-life setting and how it is not possible to maximize the positive predictive value in a similar manner to the structured training protocol. Thus, there is still a need to confirm what actually caused any event above a given threshold. During the regular training sessions that we observed, headers were relatively infrequent. But even if a greater proportion of headers would most likely yield higher positive predictive values, there would still be a need for video confirmation, for example.

As the main aim of this study was to evaluate the sensor's potential for usage in large-scale data collection in youth soccer, practical considerations on feasibility and user-friendliness need to be addressed. We encountered several software problems during the course of the study, such as having to retrieve apparently missing data from one of the on-field sessions. Furthermore, players differed in their opinions regarding whether they would accept wearing the sensors over longer periods throughout the season, including matches. Despite being designed with a lumen to minimize any hearing impairment, this seemed to be one of the main criticisms. We also observed that some of the sensors were partially obstructed with cerumen after the sessions. Such concerns are likely to limit the utility of such devices; they not only render the data potentially unreliable but might also negatively affect compliance.

We acknowledge several study limitations. First, a laboratory validation needs to rely on a reference system, with its own imperfections. We chose a well-recognized method (HIII) to make our results comparable with the work of others, as well as easy to replicate. Initially, we performed a thorough assessment of frontal impacts (considered most relevant for soccer) and then proceeded to address the issues of impact location and severity. This explains the discrepant number of impacts across conditions. We chose to exclude and replace data from a series of severe outliers. We did this because we consider the suspected cause plausible: a specific mechanical response of the HIII head and neck during a sequence of impacts gave rise to an artifact in the MV1 sensor. Such an artifact may reflect specific technical sensor characteristics, including sample rate and sensor resonant frequency response or bandwidth. Including these data would potentially disguise our main findings, as this particular issue does not reflect a challenge related to the real-life human scenarios that we ultimately evaluated. Second, we recognize that only 6 players took part in this study and that only 2 regular training sessions were included, potentially limiting the external validity to other playing levels, sex, and styles of play. Compensating for this, we have a data set composed of several hundreds of events. Last, for logistical reasons, we attached the sensors ourselves for the on-field parts of the study, without an on-site demonstration recommended by the producer. Even though this might be a source of systematic error, we did our best to comply with the instructions. In summary, however, it seems unlikely that these limitations invalidate our main findings.

The main strength of this study lies in its stepwise approach, allowing us to translate our findings from the laboratory into a real-life setting. As a result, we believe that our findings have illustrated several challenges that need to be taken into account when considering the use of such sensor systems for quantifying head impact exposures in any collision or contact sport. We suggest that new methods be evaluated carefully before being taken into use, including not only a laboratory validation but also an on-field evaluation. Future sensor systems should seek to improve technical specifications (eg, sampling rate), create algorithms better capable of filtering out spurious non–head impact events, and optimize head coupling. Until then, it is important to remain critical when interpreting data acquired from such systems and to confirm all events with secondary means of verification.

CONCLUSION

This study highlights several previously recognized challenges when attempting to quantify head impacts in contact sports with sensor systems. It also demonstrates the need for careful and systematic evaluation before being used in real-life and research settings. In-ear sensors represent a novel method for quantifying head impact characteristics in youth soccer. However, the device tested in this study displayed considerable random error and overestimated head impact exposures substantially, depending on the severity and type of impact. Despite showing excellent on-field accuracy for discriminating headings from other accelerative events in youth soccer, absolute values should be interpreted with caution, and there is a need for secondary means of verification (eg, video analysis) in real-life settings.

ACKNOWLEDGMENT

The authors are grateful to the coaching staff of Nordstrand IF and to all the players who participated in the study, including Oliver Carlmar Bøifot, Kasper Johannesen Schierning, David Kværner-Andersen, Martin Bolstad, Aleksander Krystad, Jakob Blixt Flaten, and Andreas Grymyr Simonsen. They also thank David Corless, Henrik Myhre Ihler, and Tomas Bahr Sandmo for their assistance in data collection.

REFERENCES

1. Brennan JH, Mitra B, Synnot A, et al. Accelerometers for the assessment of concussion in male athletes: a systematic review and meta-analysis. *Sports Med*. 2017;47(3):469-478.
2. Cortes N, Lincoln AE, Myer GD, et al. Video analysis verification of head impact events measured by wearable sensors. *Am J Sports Med*. 2017;45(10):2379-2387.
3. Cummiskey B, Schiffmiller D, Talavage, et al. Reliability and accuracy of helmet-mounted and head-mounted devices used to measure head accelerations. *Proceedings of the Institution of Mechanical Engineers, Part P: Journal of Sports Engineering and Technology*. 2017;231(2):144-153.
4. Koerte IK, Ertl-Wagner B, Reiser M, Zafonte R, Shenton ME. White matter integrity in the brains of professional soccer players without a symptomatic concussion. *JAMA*. 2012;308(18):1859-1861.
5. Koerte IK, Lin AP, Muehlmann M, et al. Altered neurochemistry in former professional soccer players without a history of concussion. *J Neurotrauma*. 2015;32(17):1287-1293.
6. Koerte IK, Mayinger M, Muehlmann M, et al. Cortical thinning in former professional soccer players. *Brain Imaging Behav*. 2016;10(3):792-798.
7. Koerte IK, Nichols E, Tripodis Y, et al. Impaired cognitive performance in youth athletes exposed to repetitive head impacts. *J Neurotrauma*. 2017;34(16):2389-2395.
8. Lamond LC, Caccese JB, Buckley TA, Glutting J, Kaminski TW. Linear acceleration in direct head contact across impact type, player position, and playing scenario in collegiate women's soccer players. *J Athl Train* 2018;53(2):115-121.
9. Lipton ML, Kim N, Zimmerman ME, et al. Soccer heading is associated with white matter microstructural and cognitive abnormalities. *Radiology*. 2013;268(3):850-857.
10. McIntosh AS, Patton DA, Frechede B, Pierre PA, Ferry E, Barthels T. The biomechanics of concussion in unhelmeted football players in Australia: a case-control study. *BMJ Open*. 2014;4(5):e005078.
11. Nevins D, Smith L, Kensrud J. Laboratory evaluation of wireless head impact sensor. *Procedia Eng*. 2015;112:175-179.
12. O'Connor KL, Rowson S, Duma SM, Broglio SP. Head-impact-measurement devices: a systematic review. *J Athl Train*. 2017;52(3):206-227.
13. Patton DA. A review of instrumented equipment to investigate head impacts in sport. *Appl Bionics Biomech*. 2016;2016:1-16.
14. Press JN, Rowson S. Quantifying head impact exposure in collegiate women's soccer. *Clin J Sport Med*. 2017;27(2):104-110.
15. SAE International. *Instrumentation for Impact Test—Part 1—Electronic Instrumentation. Surface Vehicle Recommended Practice*. Warrendale, PA: SAE International; 1995.
16. Sale DG. Testing strength and power. In: MacDougal JD, Wenger HA, Green HJ, eds. *Physiological Testing of the High-Performance Athlete*. 2nd ed. Champaign, IL: Human Kinetics Books; 1991:75-82.
17. Shewchenko N, Withnall C, Keown M, Gittens R, Dvorak J. Heading in football. Part 1: development of biomechanical methods to investigate head response. *Br J Sports Med*. 2005;39(suppl 1):i10-i25.
18. Siegmund GP, Guskiewicz KM, Marshall SW, DeMarco AL, Bonin SJ. Laboratory validation of two wearable sensor systems for measuring head impact severity in football players. *Ann Biomed Eng*. 2016;44(4):1257-1274.
19. Tyson AM, Duma SM, Rowson S. Laboratory evaluation of low-cost wearable sensors for measuring head impacts in sports. *J Appl Biomech*. 2018;34(4):320-326.
20. Wu LC, Nangia V, Bui K, et al. In vivo evaluation of wearable head impact sensors. *Ann Biomed Eng*. 2016;44(4):1234-1245.
21. Zhang L, Yang KH, King AI. A proposed injury threshold for mild traumatic brain injury. *J Biomech Eng*. 2004;126(2):226-236.
22. Zhang MR, Red SD, Lin AH, Patel SS, Sereno AB. Evidence of cognitive dysfunction after soccer playing with ball heading using a novel tablet-based approach. *PLoS One*. 2013;8(2):e57364.