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1 Evaluation of an In-ear Sensor System for Quantifying Head Impacts in

2 Youth Football (Soccer)

3

4 ABSTRACT

5 **Background:** Wearable sensor systems have the potential to quantify head kinematic
6 responses of head impacts in soccer. However, on-field use of sensors, e.g. accelerometers,
7 remains challenging, due to poor coupling to the head and difficulties discriminating low-
8 severity direct head impacts from inertial loading of the head from human movements, such
9 as jumping and landing.

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

11 **Study design:** Descriptive laboratory and on-field study.

12 **Methods:** First, the sensor was mounted to a Hybrid III headform (HIII) and impacted with a
13 linear impactor or a soccer ball. Peak linear acceleration (PLA), peak rotational acceleration
14 (PRA) and peak rotational velocity (PRV) were obtained from both systems; random and
15 systematic error were calculated using HIII as reference. Then, six youth soccer players wore
16 sensors and performed a structured training protocol including heading and non-heading
17 exercises; they also completed two regular soccer sessions. For each accelerative event
18 recorded, PLA, PRA and PRV outputs were compared to video recordings. Receiver
19 operating characteristic curves were used to determine the sensor's discriminatory capacity in
20 both on-field settings, determining cut-off values for predicting outcomes.

21 **Results:** For the laboratory tests, the random error was 11% for PLA, 20% for PRA and 5%
22 for PRV; the systematic error was 11%, 19% and 5%, respectively. For the structured training
23 protocol, heading events resulted in higher absolute values (PLA=15.6±11.8g) than non-
24 heading events (PLA=4.6±1.2g); the area under the curve (AUC) was 0.98 for PLA. In
25 regular training sessions, AUC was >0.99 for PLA. A 9g cut-off value yielded a positive

26 predictive value of 100% in the structured training protocol vs. 65% in regular soccer
27 sessions.

28 **Conclusion:** The in-ear sensor displayed considerable random error and overestimated head
29 impact exposure substantially. Despite showing an excellent on-field accuracy for
30 discriminating headings from other accelerative events in youth soccer, absolute values must
31 be interpreted with caution, and there is a need for secondary means of verification (e.g. video
32 analysis) in real-life settings.

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

35 **Key words:** TBI, REPETITIVE, SOCCER, SUBCONCUSSIVE, WEARABLE,
36 ACCELEROMETER

37 **INTRODUCTION**

38 Repetitive head impacts in the ‘subconcussive’ range (i.e. head impacts without immediate
39 symptoms) are common in soccer where purposeful and unprotected heading of the ball is an
40 integral part of the game. There is evidence of long-term brain structural and functional
41 alterations in soccer players.^{5,6,8,9} Moreover, recent studies suggest a potential effect on
42 cognitive function in children and adolescents during a vulnerable period of brain
43 maturation.^{7,22} However, the link between exposure to repetitive head impacts and brain
44 alterations is still controversial and remains to be elucidated. In this context, accurate
45 measurement of head impact exposure in soccer is a key challenge when investigating the
46 effect of head impact exposure on brain health.

47 Wearable sensor systems, such as accelerometers/gyroscopes, can potentially provide
48 valuable insights into the dynamics of single and repetitive head impacts. However, several
49 issues have made quantifying head impact exposure challenging, despite the multiple systems
50 currently available.^{12,13} A central issue has been poor sensor coupling with the head; methods
51 such as skin patches and skull caps are subject to relative motion between the device and the
52 skull, and therefore not able to measure head impact exposure *in vivo* accurately.²⁰ This issue
53 extends beyond erroneous outputs for direct head impacts; failure to discriminate these from
54 non-head impact accelerative events typically seen during game play (running, jumping,
55 tackling etc.) also leads to high false positive rates.^{3,14} Thus, previous studies have concluded
56 that secondary means of verification, such as video analysis, are needed to verify whether the
57 events recorded actually represent head impacts.^{2,3,14} This makes surveillance of exposure in
58 large-scale cohort studies considerably more demanding.

59 Recently, in-ear sensor systems have become commercially available, aiming to improve
60 skull coupling by custom-molded placement in the bony portion of the external ear canal.
61 However, before usage in prospective cohort studies, it is necessary to evaluate their

62 performance in both a laboratory and an on-field setting. The aim of this paper was to test the
63 validity of a new in-ear sensor for quantifying head impacts in youth soccer.

64

65 **METHODS**

66 **Study design and participants**

67 This study was conducted in three separate phases: (1) validation of the in-ear sensor in a
68 controlled laboratory setting, (2) controlled on-field evaluation of its ability to differentiate
69 headings from other accelerative events typically seen in soccer, and, finally, (3) on-field
70 evaluation in regular soccer training sessions. In phases 2 and 3, six male youth soccer players
71 (age 15.3 ± 0.3 years, height 170.3 ± 5.0 cm, mass 54.8 ± 6.1 kg) participated, all playing at the
72 regional elite youth level in Norway during the 2017 season. The Ethics committee at the X
73 institution approved the study, and written informed consent was obtained from the
74 participants and their legal guardians.

75

76 **The in-ear sensor**

77 MV1 (MV1, MVTrak, Durham, NC, USA) is a sensor system designed for custom-molded
78 placement in the left external ear canal to optimize coupling to the head. A small lumen runs
79 through the sensor to allow air conduction, limiting hearing loss to approximately 3 dB. The
80 sensor samples linear acceleration and rotational velocity data at 1000 Hz, filtering the data
81 with a phaseless 300 Hz 8-pole low-pass Butterworth filter to remove noise; rotational
82 acceleration is calculated by differentiating this filtered rotational velocity data. The sensor
83 then provides a time-stamped output of peak linear acceleration (PLA), peak rotational
84 velocity (PRV) and peak rotational acceleration (PRA) for all accelerative events exceeding
85 3 g (i.e. nominal head impacts), followed by a 250 ms latency period before another impact
86 can be registered. The sensor stores event-specific data on a microchip, and connects via USB

87 to a computer for download. Raw data are uploaded to the MVTrak server, before being
88 processed by the producer's algorithm. These data can then be downloaded for each player
89 (i.e. sensor) as time-stamped and event-specific summaries in Excel charts, including PLA,
90 PRV, PRA and the individual kinematic components of each accelerative event.

91

92 **Experimental procedures**

93 **Phase 1, Laboratory validation.** The MV1 sensor was mounted at the ear region of an in-
94 calibration Hybrid III (HIII) head and neck assembly. Three mounting configurations were
95 assessed: (1) a custom-made flat MV1 sensor (MV1 flat) attached with double-sided tape,
96 reinforced with external taping to minimize relative motion between the HIII headform and
97 the sensor, to optimize the coupling conditions and assess this as an alternative to in-ear
98 mounting; (2) a regular in-ear MV1 (MV1 in-ear) firmly placed in a tight canal on the HIII
99 headform, representing an artificial ear canal; and (3) a regular in-ear MV1 (MV1 loose)
100 loosely placed by expanding the same canal (figure 1). We created the canal by carving out a
101 piece of the artificial skin covering of the HIII headform. The tight canal's diameter was
102 slightly smaller than the sensor's, only enough to allow the compliant properties of the rubber
103 to expand and create a snug fit, mimicking real-life custom-molded placement; the expanded
104 canal's diameter was slightly larger (2-3 mm) than the sensor's, allowing slight relative motion
105 for the sensor. The HIII head was instrumented at its centre of mass with an in-calibration
106 triaxial linear accelerometer and triaxial angular velocity sensor array sampling at a rate of 20
107 kHz. Linear acceleration and angular velocity data were filtered with a SAE CC1000 filter
108 and a SAE CC180 filter,¹⁵ respectively, before computing a preliminary set of PLA and PRV
109 values for each impact. PRA values were calculated by differentiating the filtered angular
110 velocity data. HIII-measured impact characteristics were considered reference values for MV1
111 flat; for evaluating between-sensor agreement, MV1 flat was considered reference for MV1

112 in-ear. The HIII was impacted at selected locations over a range of selected magnitudes with
113 (1) a linear impactor with a stiff interface, (2) a linear impactor with a compliant interface, or
114 (3) a FIFA-approved soccer ball inflated to 11 PSI (table 1). Each test was videoed with a
115 GoPro HERO5 Black camera, recording at 240 Hz.
116



117
118 **Figure 1.** Mounting of the MV1 in-ear (left) and MV1 flat (left middle) on the Hybrid III
119 headform, and an example of a setup for right frontal impacts with the padded impactor
120 striking from a 45 degree angle (right middle). Shown to the right is a photo of the MV1 in a
121 real-life setup.

122
123 **Phase 2, Controlled on-field evaluation.** The six participants were invited to complete a
124 structured training protocol in a controlled setting twice while wearing a custom-molded MV1
125 in their left ear canal. The protocol was designed and supervised by research staff with long-
126 standing experience in soccer (author X and author Y), and consisted of five heading and six
127 non-heading exercise drills typical for soccer. Heading exercises included (1) finishing
128 headers, (2) redirection headers, (3) long direct headers, (4) short direct headers, and (5)
129 headers from in-air duels. Non-heading exercises included (1) shoulder-to-shoulder collisions,
130 (2) forceful shooting, (3) redirection running with maximal intensity, (4) short straight
131 sprinting with maximal intensity, (5) falling abruptly forward on the ground landing on out-
132 stretched arms, and (6) in-air duels without ball contact (losing the duel).

133 **Phase 3, In-training on-field evaluation.** The participants wore the sensors for two
134 regular training sessions with their team. The sessions were instructed by their regular
135 coaching staff, and included warm-up, passing and playing drills, in addition to regular play in
136 teams.

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

140

141 **Data analyses**

142 For the laboratory validation, the HIII kinematic time histories (e.g. linear acceleration) were
143 reviewed comparing with high-speed video of each test. The aim was to review the
144 preliminary PLA, PRA and PRV values for each test and to identify the peak values directly
145 related to the initial interaction between the impactor/soccer ball and the HIII headform. After
146 review by authors X and Y, a final set of HIII PLA, PRV and PRA values was determined.

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

158 For the structured training protocol (phase 2), the individual events of each exercise drill
159 were used as reference and compared to the time-stamped outputs from the sensors. If an
160 event failed to exceed the sensor's 3 g threshold, and therefore was not recorded, kinematic
161 values were set as follows to be included in later analyses: PLA=3.0 g, PRV=3.0 rad/s, and
162 PRA=200 rad/s². These values were set arbitrarily, assuming that these events involved
163 slightly lower values than the lowest magnitude events recorded from the sensor; this was
164 done to include them in the ROC analyses. For the regular training sessions (phase 3), all head
165 impacts were first identified on video to be included in the analyses; they were then compared
166 to their potential time-stamped sensor outputs. All other nominal head impact events recorded
167 by the sensors (i.e. either non-head impact accelerative events or spurious events) were then
168 classified according to video.

169 For both on-field evaluations (phases 2 and 3), mean values \pm SD for PLA, PRV and PRA
170 were calculated for (1) all head impact and (2) all non-head impact events; this was done
171 separately for the structured training protocol and regular training sessions. To test if head
172 impacts resulted in higher absolute peak values, independent-samples t-tests were used to
173 compare the means of the two event groups in both settings, using an *a priori* significance
174 level of $p < 0.05$. Then, to determine the discriminatory capacity, receiver operating
175 characteristic (ROC) curves were constructed for each dependent variable (PLA, PRA and
176 PRV) in both settings. Expressed as area under the curve (AUC), results were interpreted as
177 excellent (1.0-0.9), good (0.9-0.8), fair (0.8-0.7) or fail (0.7-0.6). To investigate how the
178 sensor would perform in settings without other verification means, sensitivity and positive
179 predictive value were then calculated in both settings according to different PLA or PRV cut-
180 off values identified from the ROC curve.

181 SPSS version 24 (IBM SPSS Statistics, IBM Corporation, Chicago, IL) was used for all
182 statistical analyses.

183 **RESULTS**

184 **Phase 1, Laboratory validation.** For MV1 flat, 112 impacts were included for final analyses
185 (table 1). When reviewing HIII outputs, we excluded angular kinematic data only (i.e. PRA
186 and PRV) from one of the 112 impacts, since we were unable to identify the appropriate
187 initial peak values. Furthermore, for one series of consecutive impacts (n=12), all within the
188 same time period with identical set-up on the same afternoon, angular kinematic values (PRA
189 and PRV) from MV1 flat were recognized as severe outliers (values ranging from four to 13
190 times higher than the reference). Upon our request, the MV1 producer reviewed the data for
191 these specific impacts, and suspected that vibrations between the MV1 flat and the HIII was
192 the cause. We replaced these data points with outputs from MV1 in-ear from the same
193 impacts.

194 As shown in figure 2, PLA values showed the strongest correlation, followed by PRV and
195 PRA. The random error for all impacts was 11% for PLA, 20% for PRA and 5% for PRV.
196 The systematic error was 11% for PLA, 19% for PRA and 5% for PRV. The random error
197 varied with impact type and location, consistently overestimating PLA, PRA and PRV values
198 (table 1). When testing for agreement between MV1 flat and MV1 in-ear for the soccer ball
199 impacts (n=29 for PLA; n=28 for PRA and PRV values), using MV1 flat as reference, the
200 random error was 6% for PLA, 20% for PRA and 6% for PRV; the systematic error was -5%
201 for PLA, -23% for PRA and -3% for PRV.

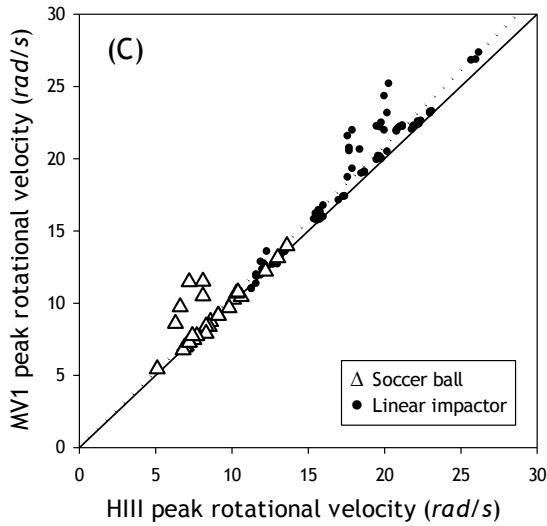
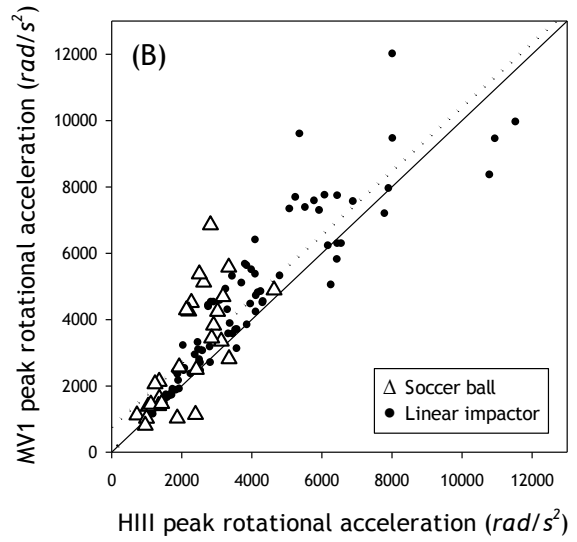
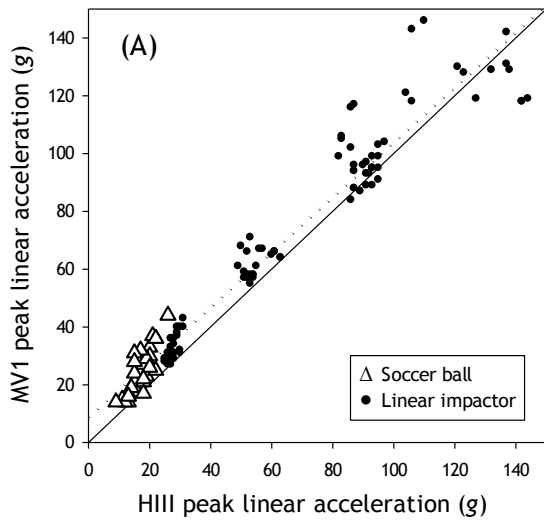
202 For MV1 loose, we replicated seven right frontal impacts and one frontal impact (HIII
203 PLA range: 29-122 g) also used for mounting configuration 2 (i.e. MV1 in-ear). Compared to
204 MV1 in-ear, the loose coupling in mounting configuration 3 led to an increase in the random
205 error from 10% to 14% for PLA, 10% to 55% for PRA, and 7% to 20% for PRV. Systematic
206 error increased from 17% to 33% for PLA, 19% to 202% for PRA, and 13% to 32% for PRV.

207 **TABLE 1.** Comparison between the reference (Hybrid III headform) and MV1 flat, with random and systematic error of PLA, PRA and PRV
 208 values, according to impact type and location.

| Impact type and location | No. of impacts | PLA range (g) | PRA range (rad/s ²) | PRV range (rad/s) | 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 - 11537 | 11 - 20 | 12 | -4 | 13 | -6 | 7 | 8 |
| Total | 83 | 25 - 144 | 1121 - 11537 | 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* | 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* | 9 - 26 | 722 - 4638 | 5 - 14 | 18 | 45 | 33 | 39 | 10 | 7 |

PLA, peak linear acceleration. PRA, peak rotational acceleration. PRV, peak rotational velocity.

*PRA and PRV values were excluded for one impact.



210

211 **Figure 2.** Peak linear acceleration (A), peak rotational acceleration (B) and peak rotational

212 velocity (C) from MV1 flat plotted against the reference (Hybrid III headform). Linear

213 regression lines (dotted) with reference lines (solid) are for all head impacts combined (i.e.

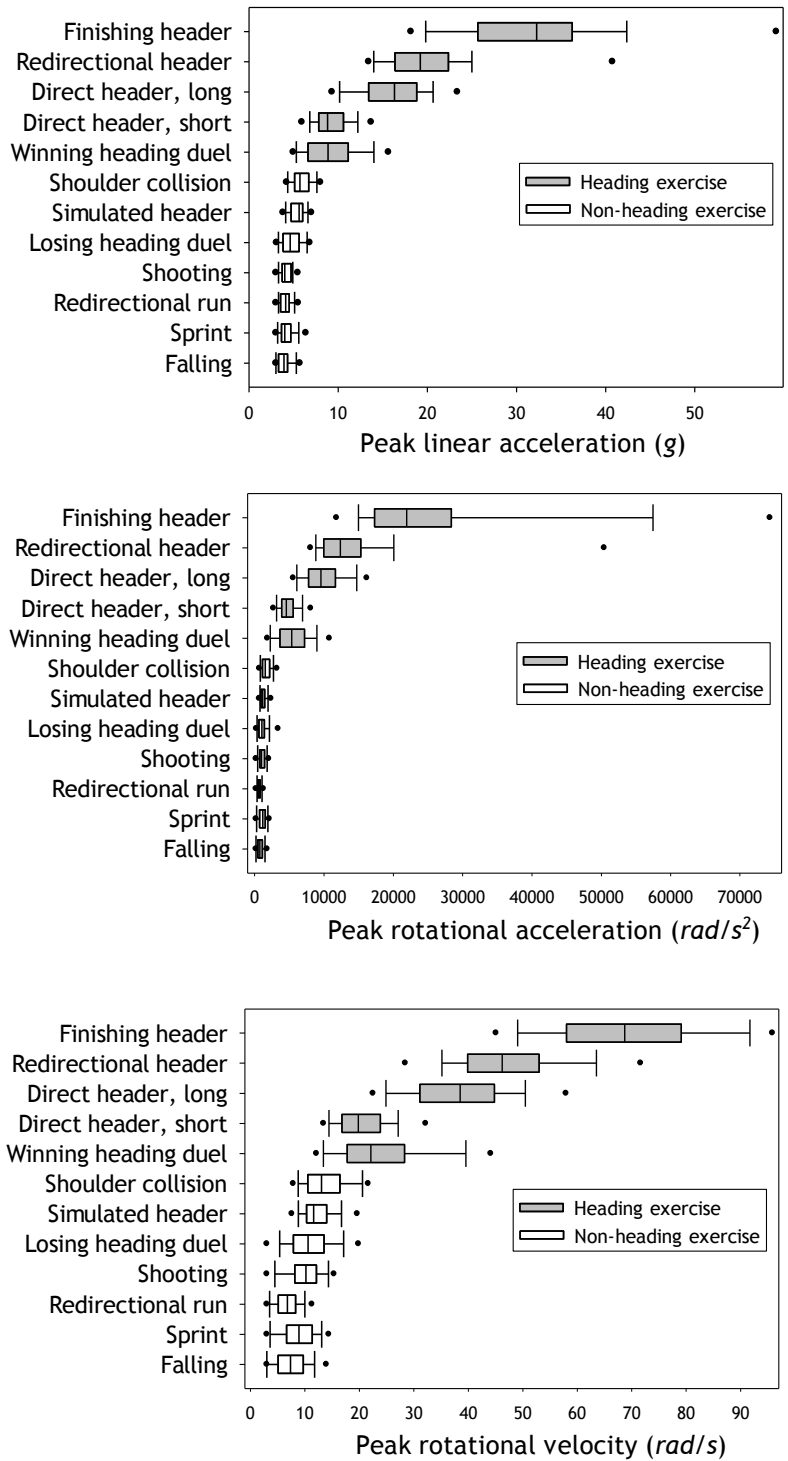
214 with linear impactor and soccer ball).

215

216

217 **Phase 2, controlled on-field evaluation.** All six participants completed each exercise
218 drill at least once, with the number of events obtained per drill ranging from 44 to 180.
219 Heading events (n=431) resulted in higher average values for all three variables
220 (PLA=15.6±11.8 g, p<0.001; PRA=10543±10854 rad/s², p<0.001; PRV=35.1±18.3 rad/s,
221 p<0.001) compared to non-heading events (n=750) (PLA=4.6±1.2 g; PRA=1095±823 rad/s²;
222 PRV=9.8±4.6 rad/s). ROC curve analyses revealed an AUC of 0.98 (95% CI 0.98 to 0.99,
223 p<0.001) for PLA, 0.99 (95% CI 0.99 to 1.00, p<0.001) for PRA and 0.97 (95% CI 0.96 to
224 0.98, p<0.001) for PRV. Figure 3 shows the distribution of peak values for each specific
225 exercise.

226 **Phase 3, in-training on-field evaluation.** Five of the participants completed one or both
227 of the regular training sessions, and, from the resulting eight sessions, the MV1 sensors
228 recorded 2 039 nominal head impact events. Of these, 15 events were confirmed on video
229 analysis to be direct head impacts (PLA=20.7±10.6 g, p<0.001; PRA=14541±7994 rad/s²,
230 p<0.001; PRV=43.5±16.4 rad/s, p<0.001), all of them due to purposeful heading of the ball.
231 No other head impacts were identified on video. The remaining 2 024 events were triggered
232 by non-head impact events such as jumping, tackling, running with change of direction,
233 touching or losing the sensor (PLA=4.0±3.1 g; PRA=835±2541 rad/s²; PRV=7.4±4.9 rad/s),
234 resulting in an AUC of >0.99 (95% CI 0.99 to 1.00, p<0.001) for both PLA, PRA and PRV.
235 Tables 2 and 3 show sensitivity and positive predictive value for different cut-off values for
236 PLA and PRV, for both the structured training protocol and the regular training sessions.



237

238 **Figure 3.** Box plots showing median value and interquartile range of peak linear acceleration,
 239 peak rotational acceleration and peak rotational velocity from MV1 for the exercises from the
 240 structured training protocol. The left and right markers are for the 5th and 95th percentile,
 241 respectively.

242

243 **TABLE 2.** MV1 sensitivity and positive predictive value for classifying accelerative events as
 244 head impacts (i.e. headers) or non-head impacts for different peak linear acceleration (g) cut-
 245 off values.

| Cut-off value (g) | Sensitivity (%) | | Positive predictive value (%) | |
|-------------------|-------------------|------------------|-------------------------------|------------------|
| | 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 |

246

247 **TABLE 3.** MV1 sensitivity and positive predictive value for classifying events as head
 248 impacts (i.e. headers) or non-head impacts for different peak rotational velocity (rad/s) cut-off
 249 values.

| Cut-off value (rad/s) | Sensitivity (%) | | Positive predictive value (%) | |
|-----------------------|-------------------|------------------|-------------------------------|------------------|
| | 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 |

250

251

252 DISCUSSION

253 This is the first study to investigate the validity of using in-ear sensors to quantify head
 254 impact exposures in youth soccer. We found that the sensor systematically overestimated head
 255 kinematic parameters and with a considerable random error (phase 1). Still, the accuracy for
 256 discriminating headers from non-head impact accelerative events in a controlled on-field
 257 setting was excellent (phase 2). However, as the proportion of head impacts (i.e. headers) was
 258 relatively low compared to non-head impact events, false positive results nevertheless
 259 remained a challenge when used in the real-life setting (phase 3).

260 Obtaining accurate results from compact, wearable sensor systems is difficult, as shown
261 by Cummiskey et al.⁴ and others^{11,18,19}. A recent review by Patton¹³ described multiple
262 examples of large discrepancies even in controlled laboratory settings. In the laboratory
263 validation (phase 1), we therefore aimed to test the technical performance of the in-ear sensor,
264 optimizing all factors, including coupling to the head. We found a consistent systematic
265 overestimation for all peak values (PLA, PRA and PRV) and with a considerable random
266 error, varying with impact type and location. Even though the exact reasons for this are
267 uncertain, several previously recognized technical limitations such as low sampling rate
268 (1 kHz for the in-ear sensor vs. 20 kHz for the reference system) might account for some of
269 the discrepancy. The observation that the PRA component generally performed poorer than
270 PLA and PRV simply reflects that it is derived from PRV, rendering it more susceptible to
271 noise and to the relatively low sample rate. This is consistent with the finding that PRA values
272 also displayed considerably poorer agreement between sensors (approx. 80%), compared to
273 both PLA and PRV (approx. 95%). As an additional barrier, algorithms of any externally
274 mounted system need to correct for its relative position on the head, in order to measure what
275 is happening at the center of mass.

276 As we were interested in how on-field conditions could affect sensor performance in
277 phases 2 and 3, we included a loose mounting configuration in phase 1. The idea was to test
278 how poor coupling could affect the inherent issues described above. With an unfavorable
279 effect on both systematic and random error for all variables, we observed a ten-fold increase
280 in the systematic error for PRA. We believe this effectively illustrates why one should
281 interpret absolute kinematic values from sensor systems in contact sports with caution. We
282 suspect that the main explanation for some of the very high on-field values observed (see
283 figure 3) is a combination of inherent systematic overestimation and poor head coupling.
284 Arguably, a mean value of well over 20 $krad/s^2$ for finishing headers almost certainly

285 represents a gross overestimation, based on previous biomechanical studies from heading in
286 soccer and mild traumatic brain injuries^{1,10,17,21}; the players considered the exercise to be in
287 the upper but normal heading severity range.

288 Press and Rowson¹⁴ recently quantified head impact exposure in collegiate women's
289 soccer using a skin patch placed behind the ear. They observed that the recorded number of
290 head impacts vastly exceeded those confirmed on video, concluding that data from head
291 impact sensors warrant careful interpretation when used in automated settings. Cortes et al.³
292 drew similar conclusions when measuring head impact exposure in lacrosse, both studies
293 highlighting the need to classify accelerative events with e.g. video analysis^{3,14}. Thus, the
294 main objective of the structured training protocol (phase 2) was to evaluate the in-ear sensor's
295 capacity to discriminate head impacts from non-head impact accelerative events. Classifying
296 all recorded accelerative events into these two main categories, in both the structured training
297 protocol and the regular training sessions, our results showed that the sensor displayed an
298 excellent discriminatory capacity. However, despite the ability to maintain high sensitivity
299 and specificity, there is a crucial difference between the two on-field settings, with real-life
300 implications. In the structured training protocol, it was possible to use a cut-off value (e.g. 9 g,
301 see table 2) yielding 100% positive predictive value, while still maintaining a sensitivity over
302 70%. In such a scenario, although missing many head impacts in the lowest range, one can
303 safely conclude that any event above this threshold is actually caused by a direct impact to the
304 head, obviating secondary means of verification (e.g. video). We were unable to replicate this
305 finding in the regular training sessions (phase 3) due to spurious non-head impact events, such
306 as touching or dropping the sensor on the ground, recording values as high as 65 and 124 g.
307 Tables 2 and 3 illustrate the difficulties of identifying a PLA or PRV cut-off value in a real-
308 life setting, and how it is not possible to maximize the positive predictive value in a similar
309 manner as for the structured training protocol. Thus, there is still a need to confirm what

310 actually caused any event above a given threshold. During the regular training sessions we
311 observed, headers were relatively infrequent. But even if a greater proportion of headers most
312 likely would yield higher positive predictive values, there would still be a need for e.g. video
313 confirmation.

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

324 We acknowledge several study limitations. First, a laboratory validation needs to rely on
325 a reference system, with its own imperfections. We chose a well-recognized method (HIII) to
326 make our results comparable to the work of others, as well as easy to replicate. Initially, we
327 performed a thorough assessment of frontal impacts (considered most relevant for soccer),
328 then proceeding to address the issues of impact location and severity. This explains the
329 discrepant number of impacts across conditions. We chose to exclude and replace data from a
330 series of severe outliers. We did this as we consider the suspected cause plausible: a specific
331 mechanical response of the HIII head and neck during a sequence of impacts gave rise to an
332 artefact in the MV1 sensor. Such an artefact may reflect specific technical sensor
333 characteristics, including sample rate and sensor resonant frequency response or bandwidth.
334 Including these data would potentially disguise our main findings, as this particular issue does

335 not reflect a challenge related to the real-life human scenarios we ultimately evaluated.
336 Second, we recognize that only six players took part in this study and that only two regular
337 training sessions were included, potentially limiting the external validity to other playing
338 levels, sex, and styles of play. Compensating for this, we have a data set comprised of several
339 hundreds of events. Last, due to logistical reasons, we attached the sensors ourselves for the
340 on-field parts of the study, without an on-site demonstration recommended by the producer.
341 Even though this might also be a source of systematic error, we did our best to comply with
342 their instructions. In summary, however, it seems unlikely that these limitations invalidate our
343 main findings.

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

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355

356 **CONCLUSION**

357 This study highlights several previously recognized challenges when attempting to quantify
358 head impacts in contact sports with sensor systems. It also demonstrates the need for careful
359 and systematic evaluation before being used in real-life and research settings. In-ear sensors
360 represent a novel method for quantifying head impact characteristics in youth soccer.

361 However, the device tested in this study displayed considerable random error and
362 overestimated head impact exposures substantially, depending on both the severity and type
363 of impact. Despite showing an excellent on-field accuracy for discriminating headings from
364 other accelerative events in youth soccer, absolute values should be interpreted with caution,
365 and there is a need for secondary means of verification (e.g. video analysis) in real-life
366 settings.

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