Sensor systems

Lightweight Wearable Headband With Flexible Hybrid Electronics for Head-Kinematic Monitoring and Mild Traumatic Brain Injury Risk Detection

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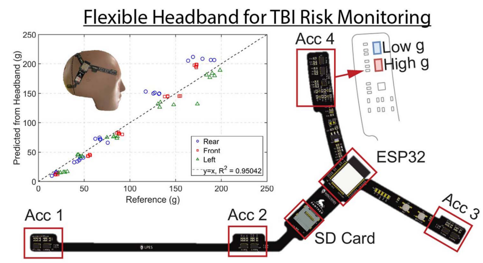
Abstract—Mild traumatic brain injuries are a significant health risk in sports and military environments, often caused by high-impact forces. This letter presents a flexible hybrid headband system for real-time monitoring of head kinematics during impacts of varying magnitude and direction. It integrates eight triaxial accelerometers in a near-regular tetrahedral configuration and employs an acceleration-only algorithm to measure linear accelerations without gyroscopes. Firmware uses a parallel queue system for efficient real-time data collection at 1600-Hz bandwidth. Testing on a Hybrid III head form via the dummy for rotational evaluation of wearable system evaluated five impact magnitudes and directions (front, rear, and left). CORrelation and Analysis (CORA) validated system accuracy, with average CORA scores of 0.840 (rear), 0.883 (front), and 0.832 (left). Some individual impacts achieved scores up to 0.98. Repeatability tests showed minimal variation, confirming consistent performance. These results demonstrate the system's potential for real-time, reliable head-kinematic monitoring in military helmets and high-impact sports.



Index Terms—Sensor systems, accelerometer, flexible hybrid electronics, head-kinematic monitoring, mild traumatic brain injury (mTBI).

I. INTRODUCTION

Mild traumatic brain injuries (mTBIs), such as concussions, are a significant health concern in the USA, affecting millions each year, with many cases likely underreported due to misdiagnosis or lack of awareness [1], [2]. In 2014, 2.87 million Americans visited emergency rooms due to TBIs [3]. TBIs in children, particularly from sports and recreational activities, contribute to approximately 283 000 emergency visits annually and account for one-third of all injury-related hospitalizations [4], [5], with youth football affecting up to 15% of players [6]. Military personnel face heightened mTBI risks, with over 400 000 cases since 2000 [7], caused by blunt impacts and blast exposures that complicate monitoring [8]. Frequent blast-related injuries also raise the risk of chronic traumatic encephalopathy (CTE): a degenerative condition associated with repeated TBIs, leading to symptoms such as memory loss, depression, and dementia [9], [10].

mTBIs often result from linear and angular head accelerations. The mechanics of how the head's motions induce mTBI is an active area of research. However, it is currently believed that an event's risk of mTBI can be captured through metrics computed from that event's time history of the head's accelerations or other kinematic measures.

Metrics such as the head injury criterion, primarily used for severe TBIs, and the brain injury criterion, developed specifically for mTBIs, provide frameworks for assessing injury severity in high-velocity impact environments, especially in sports and military settings [11], [12]. These metrics are essential for advancing injury prevention strategies in youth sports [13] and military training and exercises; enhancing the effectiveness of sports and military training; and increasing the accuracy of mTBI diagnosis and evaluation [14]. Real-time monitoring of head kinematics in sports, military training, and operations is challenging. Instrumented mouthguards (iMGs) have proven effective in high-impact sports such as football, boxing, and mixed martial arts [15], [16], [17], [18]. Alternative approaches, such as optical tracking, are impractical in dynamic environments. Wearable sensors, including in-ear and in-mouth devices, disrupt operations by impeding interpersonal communications [19], [20], [21]. Current helmet-mounted and skin-mounted devices suffer from significant errors. The head impact telemetry system (HITS) shows error rates of 18%–31% for linear acceleration and 35%–64% for rotational acceleration [22], while XPatch errors exceed 64% and 370%, respectively [23].

This letter presents a flexible hybrid electronics [24]-enabled headband system for real-time monitoring of head impacts in military and sports settings. We validate our system in ideal conditions. In the HITS, Wu et al. [25] further reported overpredictions of linear and angular accelerations by 120% and 290% due to poor skull coupling. Such limitations cannot be ruled out in real-world scenarios and in our design as well. However, despite this foreseeable limitation, the investigation and development of head-mounted sensor systems are justified due to their potential for providing nonintrusive solutions for real-time monitoring.

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Our system uses advanced accelerometers and an accelerometer-only (AO) algorithm to measure linear and angular accelerations without gyroscopes, providing precise tracking where traditional systems are less effective. [26], [27]. While this letter specifically focuses on demonstrating the validation for linear accelerations, we postulate that angular accelerations would show similar levels of accuracy based on the demonstrated agreement observed in linear measurements. Testing on a Hybrid III head form under high-impact conditions demonstrated the system's reliability and repeatability, with CORrelation and Analysis (CORA) [28] validating the system's accuracy by comparing its performance to reference sensors.

II. METHODOLOGY

A. Sensor System Design

The system uses low-g (±16 g ADXL345) and high-g (±200 g ADXL375) triaxial accelerometers with sensitivities of 4 mg/LSB and 0.049 g/LSB and resolutions of 13 bit and 16 bit, respectively. During postprocessing, it switches between accelerometers based on measured acceleration, prioritizing low-g for impacts below 16 g and high-g for higher accelerations to avoid saturation. This ensures reliable kinematic measurements across varied impact scenarios. Gyroscopes are excluded as the AO algorithm accurately predicts motion using only accelerometers, reducing power consumption and extending battery life, making the system lightweight and energy-efficient for wearable applications. All electronics are mounted on a 0.23-mm thick flexible polyamide printed circuit board (PCB) with two conductive layers. When worn, the accelerometers form a near-regular tetrahedral configuration for optimal AO algorithm performance. The system uses an ESP32 dual-core microcontroller for real-time data collection, storing data on a 32 GB secure digital (SD) card. Power is supplied via a wired connection or a 3.3-V battery.

B. Firmware Design

The firmware manages real-time data collection from all accelerometers at a sampling rate of 1600 Hz, ensuring continuous recording throughout each impact event. This high sampling rate provides precise temporal resolution, capturing detailed kinematic profiles across the entire duration of the impact rather than just at the peak moment. Such continuous data collection enables comprehensive validation and analysis, representing a marked improvement over traditional head impact sensors that focus solely on peak measurements. Accelerometers connect via serial peripheral interface (SPI), and the SD card uses SD multimediacard (SDMMC) for efficient data logging. Sessions are stored as compact *.dat* files to minimize size and enable fast read/write operations, later converted to CSV for analysis. Data collection is initiated with a button press for ease of use.

C. AO Algorithm

The AO algorithm [28], [29], developed by the Kesari Group at Brown University, enables accurate calculation of both linear and angular accelerations using only accelerometer data. Four triaxial accelerometers are arranged in a near-regular tetrahedral configuration to provide the necessary input. The AO system eliminates the need for gyroscopes, reducing power consumption and enhancing battery life for wearable applications. While obtaining precise relative positions and direction vectors adds complexity, this study used 3-D scanning during laboratory validation, providing a foundation for optimizing real-world implementation. The algorithm takes the relative positions and direction vectors of the accelerometers along with linear acceleration data to compute acceleration at any point in the system (head) and the laboratory reference frames.

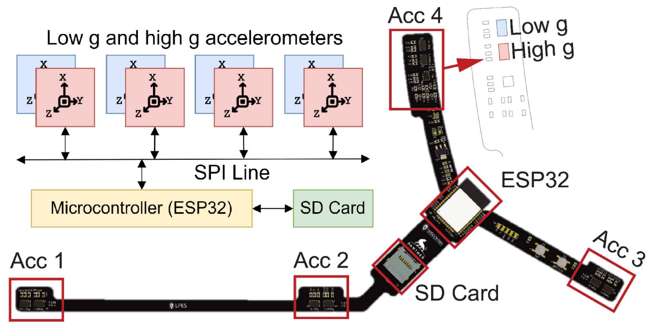

Fig. 1. Flexible polyamide PCB with mounted low-g and high-g accelerometers and the microcontroller. The block diagram shows the system architecture, with a zoomed-in view of an accelerometer location displaying the low-g and high-g sensor pair.

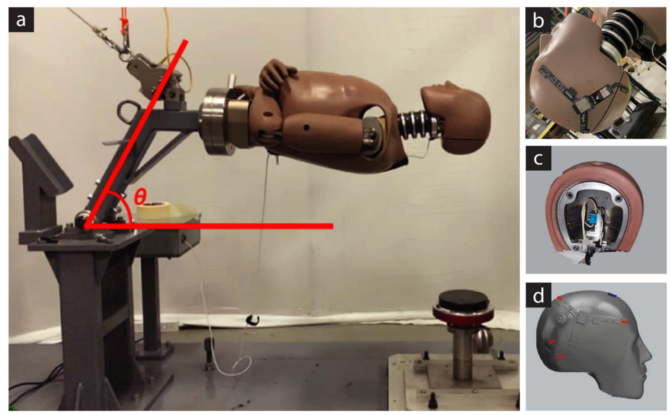

Fig. 2. (a) DREW test rig with labeled theta angle used to vary drop intensity. (b) Flexible headband mounted on the Hybrid III head form. (c) Endevco sensors placed at the CG. (d) 3-D scan of the head form with marked sensor locations.

D. Test Setup

The test setup employed the dummy for rotational evaluation of wearable (DREW) system, developed by Team Wendy in Cleveland, OH, to evaluate the accuracy of the sensor system (see Fig. 2). A Hybrid III head form, selected following the CHAMP [29] guidelines for laboratory validation of head acceleration sensors, was used as a surrogate. While the Hybrid III offers validated geometry and inertial properties, its limited biofidelity in replicating soft tissue and hair effects is noted, as highlighted in CHAMP [29]. To address these limitations, future tests will prioritize real-life conditions by incorporating biofidelic surrogates and human participants to simulate practical attachment and coupling challenges. Endevco 7264C-2K accelerometers were placed at the head's center of gravity (CG) for reference [see Fig. 2(c)]. The DREW system simulated impacts with varying accelerations across 5 g-force ranges: less than 16 g, 16–50 g, 50–100 g, 100–150 g, and 150–200 g, with five repeats for each range. Impacts were tested from the front, rear, and left under nonhelmeted conditions. These test conditions were designed to align with CHAMP recommendations for comprehensive sensor evaluation, covering a range of magnitudes,

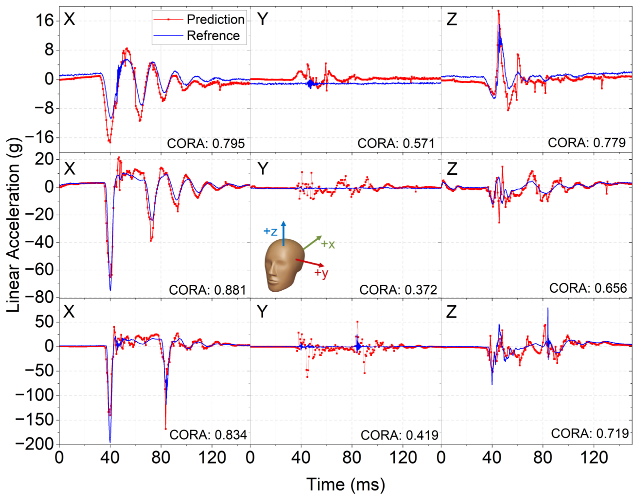

Fig. 3. Linear acceleration results for three rear impacts (low, medium, and high) in the *x, y,* and *z* directions. Each row represents an impact, and each column shows acceleration for the respective axis. The center head form illustrates the *x, y,* and *z* directions used as reference.

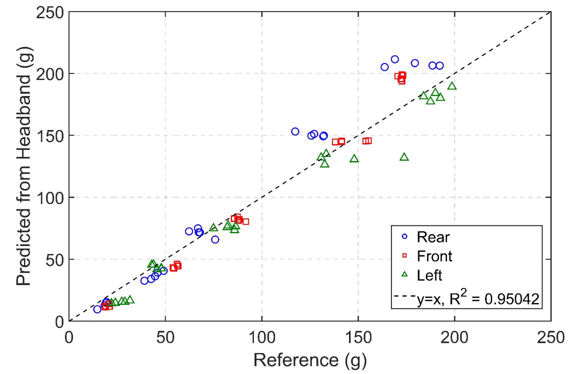

Fig. 4. Peak resultant linear acceleration for all rear, front, and left impacts, plotted against the $y = x$ line, comparing predicted and reference data.

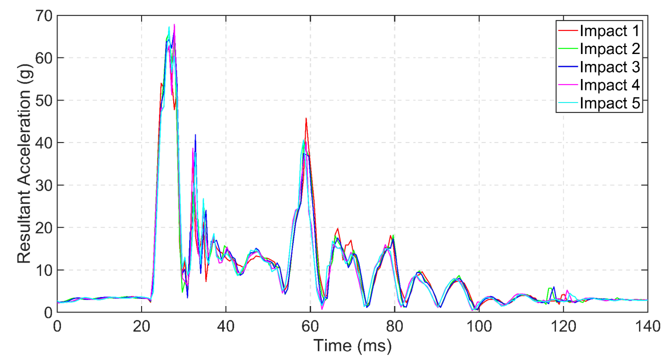

Fig. 5. Repeatability results for five rear impacts, showing consistent performance of the test setup and the sensor system across all trials.

durations, and directions. The detailed configuration of the test rig and the sensor setup is shown in Supplementary Fig. 1. After mounting, the sensors were 3-D scanned using a Handyscan 700 [see Fig. 2(d)] to extract the relative positions and direction vectors. The Handyscan 700 was used solely for precise placement validation during laboratory testing and is not intended for real-world deployments. Future work will focus on evaluating the system under practical conditions using more realistic deployment methodologies to refine its performance in field scenarios. These data were then used as input for the AO algorithm, enabling accurate comparison of the headband sensor data to the reference sensor location.

III. RESULTS

A. Linear Acceleration

The system measured linear accelerations across various impact scenarios. Fig. 3 shows three rear impacts—low (<16 g), medium (50–100 g), and high (150–200 g)—with the x-axis, the primary impact direction, correlating closely with reference data. Overall, CORA scores were 0.818 for low-g, 0.865 for medium-g, and 0.813 for high-g impacts. The x-axis recorded the highest scores, aligning with the primary direction, while lower y and z scores resulted from vibrational accelerations in the Hybrid III head form, which do not affect the DTS sensor embedded at the CG. Similar results for front and left impacts are in Supplementary Figs. 2 and 3. Fig. 4 compares predicted peak impacts using the AO algorithm with reference sensors for all directions. The R^2 values in Fig. 4 are based on a $y = x$ fit to evaluate the deviation of predicted data from an ideal 1:1 relationship with reference data. The ideal result would show all points lying on the $y = x$ line, representing perfect agreement. This approach provides a lower bound compared to R^2 derived from a calculated linear regression with a nonzero y-intercept that prescribes an offset bias.

B. Repeatability

The system's repeatability was assessed through five repeated impacts at each g-force range for rear, front, and left impacts. Fig. 5 shows the results for rear impacts, demonstrating consistent performance with minimal variation. Slight differences in impact angles, due to manual test rig adjustments, caused minor measurement variations. These results confirm the system's high repeatability.

C. CORA

CORA scores were used to evaluate the system's accuracy in predicting linear accelerations, a metric for assessing signal agreement in sensor validation studies. CORA analyzes the entire time-series data across dimensions such as phase, magnitude, and progression, providing a rating scale where 1 indicates a perfect match, and 0 represents no match. CORA averages were 0.840 for rear impacts, 0.883 for front, and 0.832 for left, and individual impacts reaching up to 0.98. Fig. 6 details rear impact scores, showing consistently high phase scores, with size and progression varying by the impact direction. Primary axes (x for rear, x and z for front, and y for left) achieved the highest scores, while nonprimary axes showed lower scores due to residual vibrations. Overall, these values are reflective of the degree of agreement between predicted and reference data across multiple impact scenarios. Similar results for front and left impacts are shown in Supplementary Figs. 4 and 5.

IV. CONCLUSION

This letter presents a novel flexible hybrid electronic headband system capable of real-time monitoring of head impacts in both military and sports environments. The headband form factor is expected to be advantageous for use adoption as opposed to iMGs due to comfort and the verbal communication requirements for both

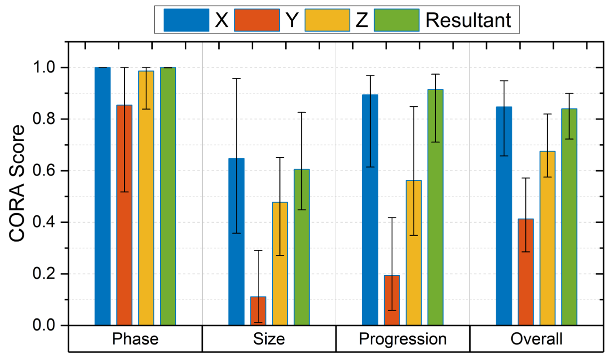

Fig. 6. CORA summary for rear impacts, showing phase, size, progression, and overall scores.

athletic and military applications. In addition, the use of multiple sensors demonstrates a higher fidelity kinematic reconstruction when compared to skin-mounted sensors, such as the XPatch, which are prone to significant errors due to relative motion with the skull. The system's use of an AO algorithm allows for accurate measurement of linear accelerations, eliminating the need for gyroscopes and reducing complexity. Testing on a Hybrid III head form yielded CORA scores of 0.840 for rear, 0.883 for front, and 0.832 for left impacts, which are representative of the degree of agreement with reference data across multiple trials. Repeatability tests confirmed consistent performance across trials. Testing on a Hybrid III head form yielded results that align with CHAMP recommendations for laboratory validation of wearable head-kinematic devices. While validation was conducted under idealized conditions, future work will include real-world testing with human participants and improved attachment methods to replicate soft tissue and hair effects, enabling a more comprehensive evaluation for deployment in high-impact sports and military applications.

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The authors Jeneel Kachhadiya, Joseph Andrews, Haneesh Kesari, and Ron Szalkowski conceived the study. Jeneel Kachhadiya and Joseph Andrews designed the system, integrating flexible embedded hardware, software, and data analysis components. Yang Wan and Haneesh Kesari contributed to the mechanical and mechanics aspects of the sensor system's design and to applying the AO algorithm to the sensor data. Jaden Romero, Shuting Kou, and Ron Szalkowski developed the Hybrid III test rig and, along with J.K. and J.A., conducted the testing. Joseph Andrews supervised the research. All authors contributed to the writing of the manuscript.

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REFERENCES

[1] J. A. Langlois, W. Rutland-Brown, and M. M. Wald, "The epidemiology and impact of traumatic brain injury: A brief overview," *J. Head Trauma Rehabil.*, vol. 21, no. 5, pp. 375–378, 2006, doi: 10.1097/00001199-200609000-00001.

[2] W. P. Meehan, R. C. Mannix, M. J. O'Brien, and M. W. Collins, "The prevalence of undiagnosed concussions in athletes," *Clin. J. Sport Med.*, vol. 23, no. 5, pp. 339–342, 2013, doi: 10.1097/JSM.0b013e318291d3b3.

[3] Centers for Disease Control and Prevention, "Traumatic brain injury-related emergency department visits, hospitalizations, and deaths," 2020. [Online]. Available: https://www.cdc.gov/TraumaticBrainInjury

[4] D. Waltzman, L. S. Womack, K. E. Thomas, and K. Sarmiento, "Trends in emergency department visits for contact sports-related traumatic brain injuries among children—United States, 2001–2018," *Morb. Mortal. Wkly. Rep.*, vol. 69, no. 27, pp. 870–874, 2020, doi: 10.15585/mmwr.mm6927a4.

[5] A. Theadom et al., "Incidence of sports-related traumatic brain injury of all severities: A systematic review," *Neuroepidemiology*, vol. 54, no. 2, pp. 192–199, 2020, doi: 10.1159/000505424.

[6] K. L. O'Connor et al., "Epidemiology of sport-related concussions in high school athletes: National athletic treatment, injury and outcomes network (NATION), 2011–2012 through 2013–2014," *J. Athl. Train.*, vol. 52, no. 3, pp. 175–185, 2017, doi: 10.4085/1062-6050-52.1.15.

[7] K. M. Lee, T. L. Khatri, and E. R. Fudge, "US department of defense warfighter brain health initiative: Maximizing performance on and off the battlefield," *J. Amer. Assoc. Nurse Pract.*, vol. 32, no. 9, pp. 720–728, 2020, doi: 10.1097/01421590040100003.

[8] E. Park, J. D. Bell, and A. J. Baker, "Traumatic brain injury: Can the consequences be stopped?," *Cmaj*, vol. 178, no. 10, pp. 1163–1170, 2008, doi: 10.1503/cmaj.080358.

[9] J. Mez et al., "Clinicopathological evaluation of chronic traumatic encephalopathy in players of American football," *J. Amer. Med. Assoc.*, vol. 318, no. 4, pp. 360–370, 2017, doi: 10.1001/jama.2017.8334.

[10] J. Bartsch et al., "Measuring blunt force head impacts in athletes," *Mil. Med.*, vol. 185, no. 1, pp. 190–196, 2020, doi: 10.1093/milmed/usz334.

[11] H. Mertz, P. Prasad, and G. Nusholtz, "Head injury risk assessment for forehead impacts," *Soc. Automot. Eng. Tech. Paper*, pp. 26–46, 1996, doi: 10.4271/960099.

[12] E. G. Takhounts et al., "Development of brain injury criteria (BrIC)," *Stapp Car Crash J.*, vol. 57, pp. 243–266, 2013, doi: 10.4271/2013-22-0010.

[13] W. T. Tsushima et al., "Incidence and risk of concussions in youth athletes: Comparisons of age, sex, concussion history, sport, and football position," *Arch. Clin. Neuropsychol.*, vol. 34, no. 1, pp. 60–69, 2019, doi: 10.1093/arclin/acy019.

[14] C. Kuo et al., "On-field deployment and validation for wearable devices," *Ann. Biomed. Eng.*, vol. 50, no. 6, pp. 1372–1388, 2022, doi: 10.1007/s10439-022-02914-3.

[15] E. E. Kiefer et al., "A two-phased approach to quantifying head impact sensor accuracy: In-laboratory and on-field assessments," *Ann. Biomed. Eng.*, vol. 48, no. 11, pp. 2613–2625, 2020, doi: 10.1007/s10439-020-02647-1.

[16] L. F. Gabler et al., "On-field performance of an instrumented mouthguard for detecting head impacts in American football," *Ann. Biomed. Eng.*, vol. 48, no. 11, pp. 2599–2612, 2020, doi: 10.1007/s10439-021-02826-8.

[17] J. Bartsch et al., "Validation of an 'intelligent mouthguard' Single event head impact dosimeter," *Stapp Car Crash J.*, vol. 58, pp. 1–27, 2014, doi: 10.4271/2014-22-0001.

[18] E. Jansen et al., "Characterizing head impact exposure in men and women during boxing and mixed martial arts," *Orthopaedic J. Sports Med.*, vol. 9, no. 12, 2021, Art. no. 23259671211059815, doi: 10.1177/23259671211059815.

[19] S. B. Sandmo et al., "Evaluation of an in-ear sensor for quantifying head impacts in youth soccer," *Amer. J. Sports Med.*, vol. 47, no. 4, pp. 974–981, 2019, doi: 10.1177/0363546519826953.

[20] A. Rezaei and L. C. Wu, "Automated soccer head impact exposure tracking using video and deep learning," *Sci. Rep.*, vol. 12, no. 1, 2022, Art. no. 9282, doi: 10.1038/s41598-022-24653-3.

[21] L. C. Wu, L. Zarnescu, V. Nangia, B. Cam, and D. B. Camarillo, "A head impact detection system using SVM classification and proximity sensing in an instrumented mouthguard," *IEEE Trans. Biomed. Eng.*, vol. 61, no. 8, pp. 2659–2668, Nov. 2014, doi: 10.1109/TBME.2014.2332325.

[22] R. Jadischke et al., "On the accuracy of the Head impact telemetry (HIT) System used in football helmets," *J. Biomech.*, vol. 46, no. 13, pp. 2310–2315, 2013, doi: 10.1016/j.jbiomech.2013.05.030.

[23] G. P. Siegmund, S. J. Bonin, J. F. Luck, and C. Bass, "Validation of a skin-mounted sensor for measuring in-vivo head impacts," in *Proc. Int. Conf. Biomech. Inj.*, 2015, pp. 182–183.

[24] Y. Khan et al., "A new frontier of printed electronics: Flexible hybrid electronics," *Adv. Mater.*, vol. 32, no. 13, pp. 1–29, 2020, doi: 10.1002/adma.202002798.

[25] L. C. Wu et al., "In vivo evaluation of wearable head impact sensors," *Ann. Biomed. Eng.*, vol. 44, pp. 1234–1245, 2016, doi: 10.1007/s10439-016-1632-8.

[26] M. M. Rahaman et al., "An accelerometer-only algorithm for determining the acceleration field of a rigid body, with application in studying the mechanics of mild traumatic brain injury," *J. Mech. Phys. Solids*, vol. 143, 2020, Art. no. 104014, doi: 10.1016/j.jmps.2020.104014.

[27] Y. Wan, A. L. Fawzi, and H. Kesari, "Determining rigid body motion from accelerometer data through the square-root of a negative semi-definite tensor, with applications in mild traumatic brain injury," *Comput. Methods Appl. Mech. Eng.*, vol. 390, 2022, Art. no. 114271, doi: 10.1016/j.cma.2021.114271.

[28] H. G. Gehre and P. Wernicke, "Objective rating of signals using test and simulation responses," in *Proc. 21st Int. Tech. Conf. Enhanc. Saf. Veh. Conf.*, 2009, pp. 1–8.

[29] L. Gabler et al., "Consensus head acceleration measurement practices (CHAMP): Laboratory validation of wearable head kinematic devices," *Ann. Biomed. Eng.*, vol. 50, no. 11, pp. 1356–1371, 2022.