

Finite Element Evaluation of Mechanical and Sensing Responses of a Helmeted Human Head under Impact

Wei-Min Chi,^{1,2} Rui-Ting Wei,³ Chao-Ming Hsu,^{3*} and Cheng-Fu Yang^{4,5**}

¹Elite Engineers College, Dongguan University of Technology, Dongguan 523808, China

²Lunghwa University of Science and Technology, Department of Electronic Engineering,
New Taipei City 333, Taiwan

³Department of Mechanical Engineering, National Kaohsiung University of Science and Technology,
Kaohsiung 807, Taiwan

⁴Department of Chemical and Materials Engineering, National University of Kaohsiung,
Kaohsiung 811, Taiwan

⁵Department of Aeronautical Engineering, Chaoyang University of Technology, Taichung 413, Taiwan

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A sensing-oriented finite element framework is developed to evaluate impact-induced mechanical responses and their relationship with acceleration-based head injury assessment. A three-dimensional helmet–head model consisting of a helmet shell, nonlinear expanded polystyrene liner, skull, and simplified brain structure is established. Dynamic impact simulations are performed for frontal, rear, top, and lateral orientations under different impact velocities. Stress, displacement, and acceleration responses are numerically extracted from the finite element model, and the acceleration–time histories are used for head injury criterion assessment. The results reveal direction-dependent stress concentrations and velocity-dependent deformation, with frontal and rear impacts generally producing higher local stresses. Increasing impact velocity also results in greater rebound displacement and more severe dynamic responses. Beyond the conventional mechanical response analysis, the proposed framework relates spatial stress and displacement characteristics to the numerically obtained acceleration–time responses of the simulated head, which represent the physical quantities that could subsequently be measured using accelerometers or inertial measurement units in an instrumented helmet. These responses also provide reference information for determining potential sensing locations and measurement requirements.

1. Introduction

Motorcycle transportation remains one of the most prevalent commuting methods in densely populated regions, particularly in urban environments where mobility efficiency and convenience are critical. However, the extensive use of motorcycles is accompanied by a high

*Corresponding author: e-mail: jammy@nkust.edu.tw

**Corresponding author: e-mail: cfyang@nuk.edu.tw

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frequency of traffic accidents, in which head injuries are consistently identified as the leading cause of fatality and severe disability. The human head is highly vulnerable to impact loading owing to its complex anatomical structure and limited tolerance to rapid acceleration. As a result, effective protective equipment is essential for mitigating injury severity. Helmets serve as the primary protective system for motorcyclists, and their performance plays a decisive role in reducing impact-induced trauma. Therefore, understanding the biomechanical response of the head under impact conditions is crucial for improving safety standards and enhancing injury prevention strategies, particularly under varying impact scenarios. Among the various injury assessment metrics, the head injury criterion (HIC) has been widely adopted as a standard quantitative indicator for evaluating the likelihood and severity of head injury.^(1,2)

HIC is calculated from the acceleration–time history and incorporates both the magnitude and duration of the acceleration pulse. Compared with peak acceleration alone, HIC provides a more comprehensive evaluation of injury risk by capturing the temporal characteristics of dynamic loading. This makes it particularly suitable for assessing complex impact events involving transient responses and varying loading conditions, including those with rapid-onset and short-duration pulses. Consequently, HIC has been extensively applied in automotive safety, helmet testing standards, and regulatory frameworks as a key indicator for head injury assessment. However, the accuracy of HIC evaluation is highly dependent on the reliability of the underlying dynamic response data and the correct interpretation of signal characteristics, especially when multiple determining factors such as geometry and material nonlinearity are involved. From a sensing perspective, acceleration can be regarded as a measurable signal that connects structural mechanics with injury assessment. During impact, mechanical energy is transmitted through the contacting interface and propagates within the head structure, generating acceleration responses that reflect the severity of loading conditions.⁽³⁾

The characteristics of these signals are affected by impact velocity, direction, structural geometry, and material behavior, all of which contribute to variations in peak values and signal duration. In unprotected head impacts, the absence of energy-absorbing mechanisms leads to sharper and more impulsive responses, which are strongly associated with higher injury risk. In addition, localized stress concentration and rapid deformation can further amplify acceleration signals, increasing the likelihood of severe injury. Therefore, analyzing how mechanical energy is transformed into measurable signals is essential for understanding injury mechanisms and improving predictive accuracy. Finite element analysis (FEA) has long been employed to investigate helmet–head impact biomechanics because it enables the detailed evaluation of stress distribution, deformation, acceleration, and injury-related parameters that are difficult to obtain directly from internal anatomical structures.^(4–10) Numerical investigations have covered helmeted and non-helmeted head impacts, different impact locations and velocities, HIC and acceleration responses, and the effect of energy-absorbing liner structures on head protection.^(7–10) More recent investigations have further combined computational helmet–head models with head kinematics and biomechanical injury metrics to evaluate protective performance under various impact conditions.^(11,12) In our recently published work in *Sensors and Materials*, a sensor-oriented finite element model was employed to investigate the mechanical response of a

helmeted human head, with emphasis on stress distribution, displacement, and the relationship between impact conditions and measurable acceleration responses.⁽¹³⁾ The importance of FEA in helmet-impact analysis has been well recognized, whereas the connection among spatial mechanical responses, acceleration-based injury assessment, and sensing implementation requires further clarification.

Previous studies have focused on helmet performance and head injury mechanisms under various impact conditions; however, many works emphasize mechanical responses without fully linking them to injury metrics such as HIC or interpreting the results from a sensing perspective. A comprehensive evaluation framework integrating finite element simulation with injury-related response analysis is therefore necessary. A three-dimensional head model is employed to simulate impact responses under different velocities and orientations. Particular attention is given to stress concentration and displacement behavior, as these factors directly reflect load transfer and energy distribution during impact. The results indicate that stress is primarily concentrated at the contact interface, with local amplification caused by skull geometry, while displacement responses reveal rebound behavior and increased deformation with higher impact velocities. In addition, the absence of secondary loading peaks highlights the direct nature of energy transfer in unprotected impacts, indicating limited energy dissipation.

By analyzing stress and displacement characteristics, the dynamic response of the head can be more clearly interpreted, providing a direct link between mechanical behavior and injury potential. These insights contribute to improved injury prediction and more accurate HIC evaluation, and support the development of more effective protective systems. In our previous work, the primary emphasis was placed on the mechanical protective response of the helmet–head system under impact loading. In contrast, the present work shifts the focus toward a sensing-oriented interpretation of the dynamic response by linking spatial stress and displacement distributions with acceleration-based HIC assessment. The resulting mechanical fields are further used to identify mechanically sensitive regions and to provide reference information for potential sensor placement and measurement requirements. Thus, the principal difference lies in the transition from conventional mechanical performance evaluation toward a numerical framework supporting sensor-assisted impact monitoring and injury assessment. The novelty of the present study does not lie solely in reproducing the established relationships among impact velocity, stress, deformation, and injury severity. Instead, a sensing-oriented numerical framework is introduced to connect spatial mechanical responses with measurable time-dependent signals and HIC-based injury assessment. Stress and displacement distributions are used to identify mechanically sensitive regions, while the corresponding acceleration histories provide the signal basis for injury evaluation. In this manner, the finite element results can serve as reference information for determining sensor placement, measurement range, and expected signal characteristics in future instrumented helmets. This framework therefore extends the conventional helmet impact simulation toward sensor-assisted impact monitoring and injury assessment.

2. Experimental Procedure

A three-dimensional finite element model of a helmeted human head was established to evaluate impact response and HIC.⁽⁷⁾ The model includes a helmet shell, an expanded polystyrene (EPS) liner, and a human head composed of skull and simplified brain structures, as illustrated in Fig. 1, including (a) the helmet and energy-absorbing liner model, (b) the skull model, (c) the overall model, and (d) a cross-sectional view of the overall model.⁽¹⁴⁾ The helmet–head assembly was configured with appropriate contact interfaces to ensure realistic load transfer from the external impact surface through the helmet to the head. This integrated setup enables the simulation of stress propagation, deformation, and acceleration responses that are directly related to injury evaluation. In particular, the stress distribution within the skull reflects the local load transfer characteristics, while displacement behavior represents the global deformation and rebound response of the head during impact, both of which are critical for interpreting injury mechanisms. Material properties were assigned according to the functional roles of each component. The helmet shell and skull were modeled as elastic materials, while the brain was defined with lower stiffness to represent its deformable behavior. The EPS liner was modeled using a crushable foam constitutive law to capture nonlinear compression and energy absorption. This nonlinear behavior is critical because it governs the attenuation of impact energy and directly affects the amplitude and duration of acceleration signals used for HIC calculation. At the same time, it affects how stress is transmitted to the skull and how displacement evolves during impact, thereby affecting both local stress concentration and overall head motion.

The finite element mesh was generated for all components, with finer mesh applied at contact regions such as the helmet shell, EPS liner, and skull interface. Suitable solid elements were selected to ensure numerical stability under dynamic loading. A mesh convergence analysis was conducted to verify that further refinement would not significantly alter stress, displacement, or acceleration results, ensuring reliable HIC evaluation, as illustrated in Fig. 2, including (a) the meshed helmet and energy-absorbing liner model, (b) the meshed skull model, and (c) the meshed overall model. This procedure is particularly important for accurately capturing stress gradients and displacement responses, which directly affect the interpretation of impact severity.

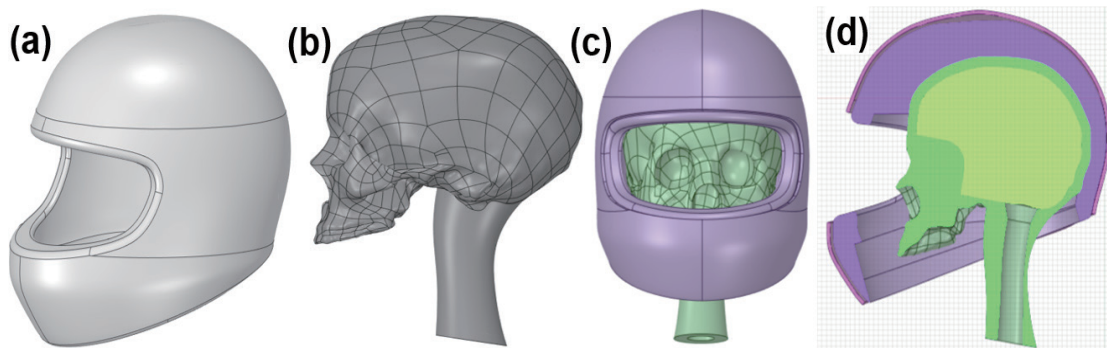


Fig. 1. (Color online) (a) Helmet and energy-absorbing liner model, (b) skull model, (c) overall model, and (d) cross-sectional view of the overall model.

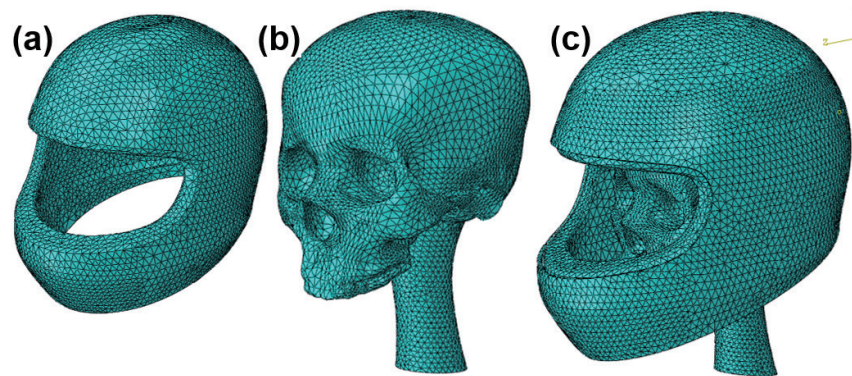


Fig. 2. (Color online) (a) Meshed helmet and energy-absorbing liner model, (b) meshed skull model, and (c) meshed overall model.

Dynamic impact simulations were performed using explicit time integration. A rigid flat surface was defined as the impact target, and the helmeted head model was assigned initial velocities toward the surface to simulate impact. Four impact orientations, frontal, rear, top, and lateral, were considered to evaluate directional effects on mechanical response and injury risk. Contact conditions were defined between the helmet and the rigid surface, as well as between the helmet and the head, to ensure realistic interaction, as illustrated in Fig. 3, including (a) frontal impact, (b) rear impact, (c) top impact, and (d) lateral impact.

In addition, modal analysis was conducted to identify the fundamental dynamic characteristics of the helmet–head system. During each dynamic simulation, stress and displacement time histories were numerically extracted from the finite element nodes/elements at the predefined impact regions corresponding to the frontal, rear, top, and lateral cases. The curves presented in Figs. 4–7 were generated directly from these simulation outputs under the specified impact velocities and therefore represent the authors’ own numerical results rather than the data obtained from external sources or experimental measurements. The acceleration–time response of the simulated head was also numerically obtained and used to evaluate HIC by considering the magnitude and duration of acceleration. No physical sensors or measurement devices were used to acquire the results presented in Figs. 4–7.

From a sensing implementation perspective, the simulated acceleration response corresponds to the quantity that could subsequently be measured using a tri-axial accelerometer or inertial measurement unit (IMU) positioned near the head center-of-gravity region, while local impact loading and liner deformation could be monitored using thin-film pressure or strain sensors embedded in the helmet liner. Thus, the numerically obtained response histories provide reference information for the future selection of sensor locations, measurement ranges, sampling requirements, and signal characteristics in experimental helmet-impact measurements. The HIC value was calculated from the acceleration–time history by considering both the magnitude and duration of acceleration. By comparing stress distribution, displacement evolution, and acceleration characteristics across different impact directions and velocities, the relationship between structural response and injury risk was established. This integrated analysis provides a comprehensive understanding of how impact energy is transferred, distributed, and transformed within the helmet–head system, offering valuable insights for both mechanical evaluation and sensing-based injury assessment.

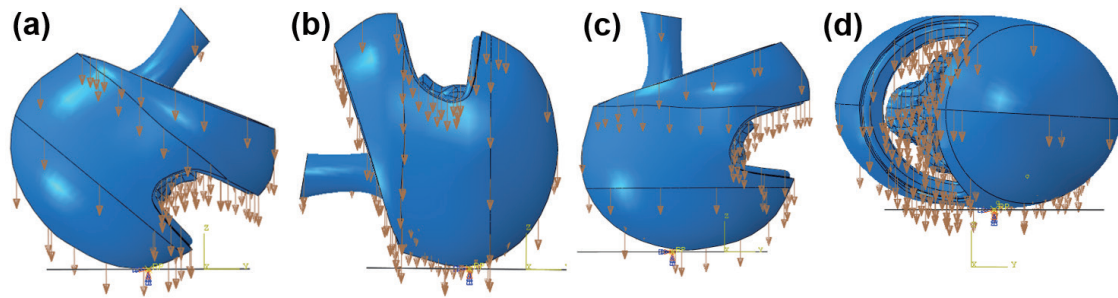


Fig. 3. (Color online) Boundary conditions applied to the overall model: (a) frontal impact, (b) rear impact, (c) top impact, and (d) lateral impact.

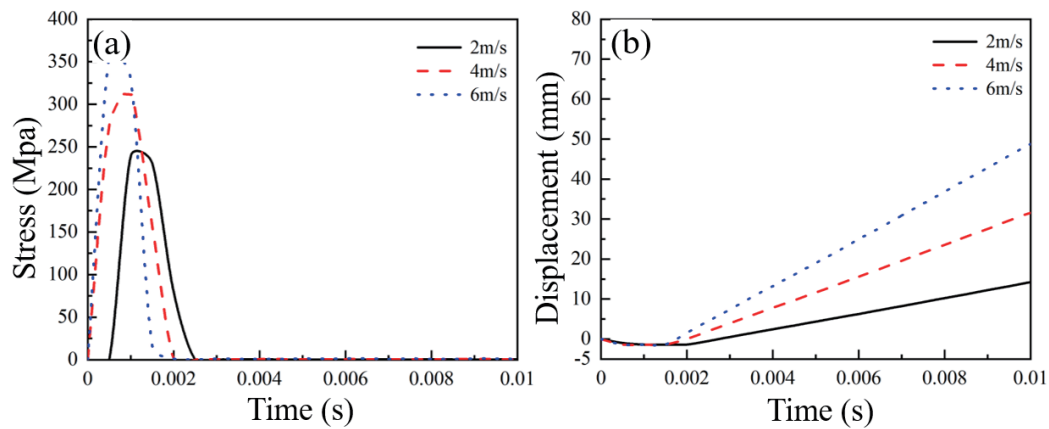


Fig. 4. (Color online) Time histories of stress and displacement at the frontal impact point of the head under varying impact parameters: (a) equivalent stress and (b) Z-axis displacement at the impact point.

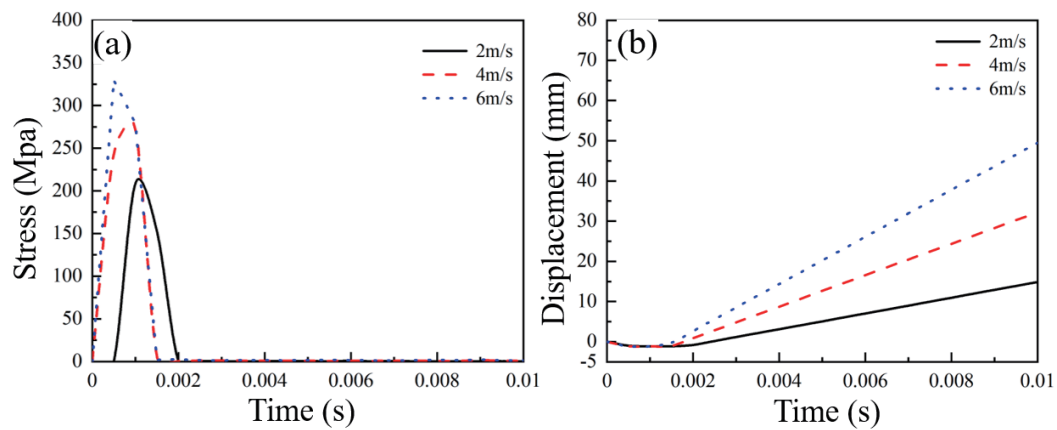


Fig. 5. (Color online) Time histories of stress and displacement at the rear impact point of the head under varying impact parameters: (a) equivalent stress and (b) Z-axis displacement at the impact point.

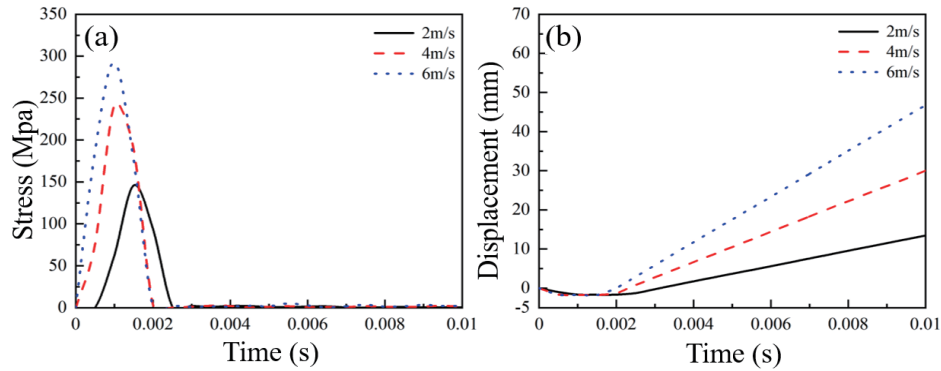


Fig. 6. (Color online) Time histories of stress and displacement at the top impact point of the head under varying impact parameters: (a) equivalent stress and (b) Z-axis displacement at the impact point.

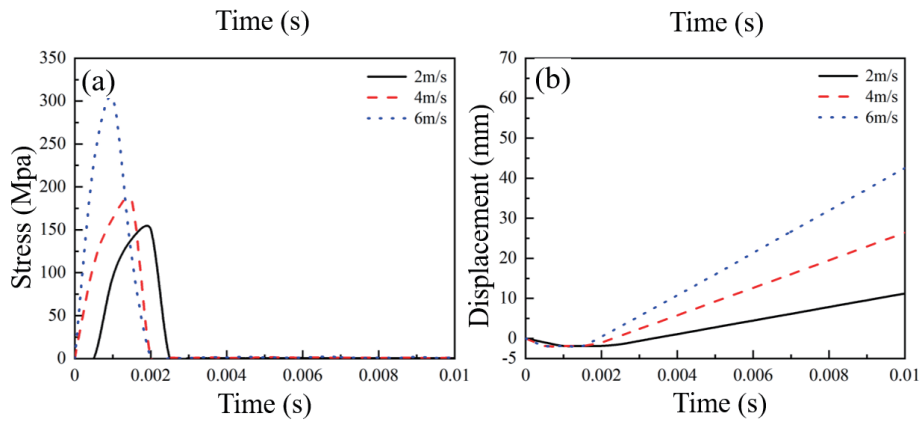


Fig. 7. (Color online) Time histories of stress and displacement at the lateral impact point of the head under varying impact parameters: (a) equivalent stress and (b) Z-axis displacement at the impact point.

3. Results and Discussion

The head impact responses are analyzed on the basis of the defined impact parameters, focusing on stress and displacement behavior. The data acquisition point for the frontal head impact is defined at the contact region in Fig. 3(a), where the maximum stress is concentrated at the interface between the skull and the rigid surface, indicating direct load transfer at the impact location. Figure 4(a) shows the equivalent stress–time histories at the frontal head impact point under different impact velocities. The results indicate that contact occurs at approximately 0.5 ms, as reflected by the sharp rise in stress. The peak stress values, corresponding to increasing impact velocities, are 238.5, 310.8, and 355.3 MPa, demonstrating a clear increase in stress with higher impact velocity. The curve corresponding to 2 m/s shows a slightly delayed impact time compared with the higher velocities, reflecting the effect of initial velocity on contact timing. Unlike the helmet impact cases, no secondary stress peak is observed after the initial impact. This suggests that, in the absence of the helmet structure, the head undergoes a single, direct

impact without significant rebound-induced re-contact. From a mechanical perspective, this indicates a more abrupt transfer of energy with limited attenuation. From a sensing perspective, the absence of secondary peaks implies a sharper and more impulsive acceleration signal, which is typically associated with higher injury risk and higher HIC values.

Figure 4(b) presents the Z-axis displacement–time responses at the same location. A negative displacement is observed prior to 0.5 ms, representing the approach of the head toward the rigid surface. After impact, the displacement initially changes only slightly, followed by a gradual increase after approximately 2 ms, indicating rebound motion away from the rigid surface. At 10 ms, the displacement values reach 14.2, 31.6, and 48.7 mm for increasing impact velocities. These results confirm that higher impact velocities result in greater deformation and rebound displacement. The larger displacement indicates higher energy transfer and reduced capacity for energy absorption in the absence of a protective structure. The results demonstrate that direct head impact leads to significantly higher stress levels and more abrupt dynamic responses than helmeted conditions. The stress increases markedly with impact velocity, while the displacement results indicate greater rebound behavior as velocity increases. From an injury evaluation perspective, these characteristics suggest that unprotected impacts generate higher acceleration peaks and more severe injury potential. The findings highlight the critical importance of helmet protection in reducing stress transmission, moderating dynamic response, and ultimately lowering HIC values.

The data acquisition point for the rear head impact is defined at the contact region in Fig. 3(b), where stress is concentrated at the interface between the skull and the rigid surface. Two regions of higher stress are observed, which can be attributed to the non-smooth geometry of the human skull. Specifically, local protrusions on the skull surface lead to localized stress concentration, indicating that anatomical features play an important role in determining impact response. Figure 5(a) shows the equivalent stress–time histories at the rear impact point under different impact velocities. Contact occurs at approximately 0.5 ms, as indicated by the sharp increase in stress. The peak stress values, corresponding to increasing velocities, are 208.1, 274.2, and 329.5 MPa, demonstrating a clear dependence on impact velocity. The curve corresponding to 2 m/s exhibits a slightly delayed contact time compared with the higher velocities, reflecting the effect of initial velocity on impact timing. Unlike the helmet impact cases, no secondary peak is observed after the initial impact, indicating that the head undergoes a single direct loading event without significant rebound-induced re-contact. This behavior suggests a more abrupt transfer of impact energy and limited attenuation capability. From a sensing perspective, such a response would generate a sharp acceleration pulse with high peak values, which is strongly associated with increased HIC and injury risk.

Figure 5(b) presents the Z-axis displacement–time responses for the rear impact. A negative displacement is observed prior to 0.5 ms, representing the approach phase before contact with the rigid surface. After impact, the displacement changes only slightly, followed by a gradual increase after approximately 2 ms, indicating the rebound motion of the head away from the rigid surface. At 10 ms, the displacement values reach 14.9, 32.2, and 49.5 mm for increasing impact velocities. These results indicate that higher impact velocities result in greater deformation and rebound displacement, reflecting increased energy transfer during impact. The

relatively large displacement values also suggest that, in the absence of a protective structure, the head experiences significant motion after impact, which may contribute to additional injury mechanisms such as rotational effects or secondary impacts. The results demonstrate that rear head impact produces high stress levels and rapid dynamic responses, with stress magnitude strongly affected by both impact velocity and local skull geometry. The absence of secondary peaks and the presence of sharp stress pulses indicate a direct and unmitigated impact process. Combined with the displacement results, these findings suggest that unprotected head impacts lead to higher acceleration peaks and increased HIC values, further emphasizing the importance of helmet protection in reducing injury risk.

The data acquisition point for the top head impact is defined at the contact region in Fig. 3(c), where stress is concentrated at the interface between the skull and the rigid surface, with higher stress observed at local protrusions owing to the non-smooth skull geometry. Figure 6(a) shows the equivalent stress–time histories at the top impact point under different impact velocities. Contact occurs at approximately 0.5 ms, as indicated by the rapid increase in stress. The peak stress values, corresponding to increasing velocities, are 145.5, 241.3, and 292.5 MPa, demonstrating a clear velocity-dependent response. Compared with the frontal and rear impact cases, the peak stress at lower velocity is relatively smaller, but increases significantly at higher velocities, indicating that the top impact is more sensitive to velocity changes. The stress curves exhibit a single dominant peak without noticeable secondary peaks, suggesting a relatively stable contact condition with minimal rebound-induced re-contact. From a mechanical perspective, this indicates a more uniform load transfer along the primary structural axis of the head. From a sensing viewpoint, such a response produces a smoother acceleration signal with fewer fluctuations, although high peak values at higher velocities may still contribute to significant HIC values.

Figure 6(b) presents the Z-axis displacement–time responses for the top impact. A negative displacement is observed prior to 0.5 ms, representing the approach phase before contact with the rigid surface. After impact, the displacement initially remains relatively stable due to the absence of energy absorption mechanisms, followed by a gradual increase after approximately 2 ms as the head rebounds and moves away from the rigid surface. At 10 ms, the displacement values reach 13.72, 29.9, and 46.7 mm for increasing impact velocities. These results indicate that higher impact velocities result in greater deformation and rebound displacement, reflecting increased energy transfer during impact. The top head impact exhibits a relatively uniform stress distribution and a stable temporal response compared with other impact directions. However, the stress magnitude still increases significantly with velocity, and the absence of attenuation mechanisms results in the direct transmission of impact energy. The displacement results further confirm pronounced rebound behavior at higher velocities. From an injury assessment perspective, these characteristics suggest that top impacts can still produce high acceleration peaks and elevated HIC values, particularly under high-speed conditions.

The lateral head impact data point is defined at the contact region in Fig. 3(d), where stress is concentrated at the interface between the skull and the rigid surface, with higher stress at local protrusions owing to the skull's non-uniform geometry. Figure 7(a) shows the equivalent stress–time histories at the lateral impact point under different impact velocities. Contact occurs at

approximately 0.5 ms, as indicated by the rapid rise in stress. The peak stress values, corresponding to increasing velocities, are 94.4, 184.5, and 303.4 MPa, demonstrating a strong dependence on impact velocity. The curve for 2 m/s shows a slightly delayed contact time compared with higher velocities, reflecting differences in initial momentum. Similar to other unhelmeted head impacts, no secondary stress peak is observed, indicating that the response is dominated by a single direct impact without significant rebound-induced re-contact. From a mechanical standpoint, this suggests that energy is transferred rapidly and with minimal attenuation. From a sensing perspective, the resulting acceleration signal is expected to exhibit a sharp peak with limited temporal spreading, which is typically associated with higher injury risk and elevated HIC values.

Figure 7(b) presents the Z-axis displacement–time responses for the lateral impact. A negative displacement is observed prior to 0.5 ms, representing the approach phase before contact. After impact, the displacement remains relatively stable for a short duration, followed by a gradual increase after approximately 2 ms, indicating the rebound motion of the head away from the rigid surface. At 10 ms, the displacement values reach 11.2, 26.4, and 42.5 mm for increasing impact velocities. These results confirm that higher impact velocities lead to greater deformation and rebound displacement, reflecting increased energy transfer during impact. The lateral head impact exhibits lower stress at lower velocities than other impact directions, but shows a rapid increase in stress with velocity, particularly at higher speeds. The presence of localized stress concentration due to skull geometry further highlights potential regions of vulnerability. The displacement results indicate consistent rebound behavior, with magnitude increasing with velocity.

From an injury evaluation perspective, these characteristics suggest that lateral impacts can still produce significant acceleration peaks and high HIC values, especially under high-speed conditions, emphasizing the importance of protective mechanisms to mitigate such effects. A sensing-oriented comparison of the four impact directions further reveals that the mechanically sensitive regions and temporal response characteristics are direction-dependent. Frontal and rear impacts produce relatively higher local stress responses, whereas top and lateral impacts exhibit different stress distributions and temporal characteristics. These differences are relevant not only to mechanical injury assessment but also to sensor implementation, because sensor location and measurement range should account for impact direction and local response magnitude. Accordingly, the new finding is the identification of direction-dependent mechanical responses as practical reference information for sensing-location selection and acceleration-signal interpretation, rather than merely confirming the established relationship between impact velocity and injury severity.

4. Conclusions

The impact responses of the human head under different loading conditions were systematically investigated, focusing on stress distribution and displacement behavior as key indicators for HIC evaluation. The results demonstrate that, in all impact scenarios, stress is primarily concentrated at the contact interface between the skull and the rigid surface, with

additional local stress amplification caused by anatomical irregularities of the skull. The magnitude of stress increases significantly with impact velocity, while the absence of secondary stress peaks indicates a direct and unmitigated energy transfer process in unprotected head impacts. The displacement results consistently show rebound behavior following impact, with higher velocities leading to greater deformation and larger rebound distances. This reflects increased energy transfer and limited energy absorption capacity in the absence of protective structures. Differences among impact directions were also observed, with frontal and rear impacts producing higher stress levels, while top and lateral impacts exhibited relatively more uniform or distributed responses but still showed significant increases at higher velocities. From an injury assessment perspective, the sharp stress peaks and abrupt dynamic responses suggest the generation of high-magnitude, short-duration acceleration signals, which are closely associated with elevated HIC values and increased injury risk. The results clearly indicate that direct head impact leads to severe mechanical loading conditions and the insufficient attenuation of impact energy. These findings highlight the critical role of protective systems, such as helmets, in reducing stress transmission, moderating dynamic response, and lowering HIC values. The study provides a clear linkage between mechanical response characteristics and injury metrics, offering valuable insights for improving injury prediction and guiding the design of more effective head protection systems.

Accordingly, the contribution of this study extends beyond the established observation that higher impact velocity produces greater mechanical responses. The proposed framework connects simulated stress and displacement fields with acceleration-based injury assessment, and provides reference information for identifying mechanically sensitive sensing locations and appropriate measurement ranges. These results can support the future integration of accelerometers, pressure sensors, and flexible strain sensors into helmet shells or EPS liners. The framework may also facilitate the evaluation of sensing-layer and energy-absorbing materials for instrumented helmets that combine mechanical protection with impact monitoring and injury assessment. A limitation of the current numerical framework is the absence of experimental validation under real-world helmet-impact conditions. Therefore, the predicted stress, displacement, acceleration, and HIC responses should presently be interpreted as numerical results rather than experimentally validated measurements. Future work will employ an instrumented helmet/headform impact setup incorporating accelerometers or IMUs and pressure or strain sensors. Experimental acceleration histories and impact responses will be compared with the numerical predictions under matched impact velocities and orientations to validate the model and assess its applicability to practical helmet-impact monitoring.

Acknowledgments

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References

- 1 R. M Greenwald, J. T Gwin, J. J Chu, and J. J Crisco: *Neurosurgery* **62** (2008) 789.
- 2 G. V. Mariotti, S. Golfo, V. Nigrelli, and F. Carollo: *Am. J. Biomed. Sci. Res.* **5** (2019) 406.
- 3 <https://www.hseblog.com/understanding-safety-helmet-standards/> (accessed July 2023).
- 4 A. Madhukar and M. Ostoj-Starzewski: *Ann. Biomed. Eng.* **47** (2019) 1832.
- 5 X. Deng, Z. Du, H. Feng, S. Wang, H. Luo, and Y. Liu: *Bioengineering* **23** (2022) 723.
- 6 T. V. Minh and T. T. Tran: *Eng. Res. Express* **7** (2025) 045430.
- 7 I. Levadnyi, J. Awrejcewicz, Y. Zhang, M. F. Goethel, and Y. Gu: *J. Med. Biol. Eng.* **38** (2017) 587.
- 8 S. Meng, A. Cernicchi, S. Kleiven, and P. Halldin: *Accid. Anal. Prev.* **134** (2020) 105297.
- 9 M. Nasim, M. J. Hasan, and U. Galvanetto: *Int. J. Mech. Sci.* **233** (2022) 107673.
- 10 N. J. Cecchi, H. V. Alizadeh, Y. Liu, and D. B. Camarillo: *Front. Bioeng. Biotechnol.* **11** (2023) 1160387.
- 11 M. Nasim, S. Spadoni, P. G. Pavan, V. Brasco, and U. Galvanetto: *J. Biomech.* **176** (2024) 112362.
- 12 M. Arnesen, S. Hallstrom, P. Halldin, and A. Kulachenko: *Results Eng.* **23** (2024) 102685.
- 13 W. M. Chi, R. T. Wei, C. M. Hsu, and C. F. Yang: *Sens. Mater.* **38** (2026) 4195.
- 14 M. Arnesen, S. Hallstrom, P. Halldin, and A. Kulachenko: *Results Eng.* **23** (2024) 102685.