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A Review of Blunt Head Injury Mechanisms and Target Testing Studies with Ballistic Impact Helmet Protection

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Abstract

With the rise in military conflicts and law enforcement needs, there is a growing demand for head protection equipment performance, particularly in terms of protection efficacy against non-penetrating blunt head injuries from ballistic impact. This paper provides a summary of recent advancements in research on the mechanisms and target testing of blunt head injuries under ballistic impact helmet protection. It begins by examining the mechanisms of ballistic head blunt impact injuries, including shock wave transmission, skull-brain tissue interaction, and resulting biological responses, laying a scientific foundation for understanding blunt head impact injuries. To accurately assess the severity and risk of blunt head impact injuries, the paper introduces evaluation criteria and explores their applicability in assessing the effectiveness of ballistic impact helmets in blunt impact injury prevention. Building on this, the paper delves into the role of target testing in evaluating helmet protection effectiveness and blunt head injury mechanisms, analyzing the strengths and limitations of different target tests in simulating real ballistic impact scenarios, evaluating helmet performance, and uncovering blunt head injury mechanisms. It also anticipates the future development direction of target testing technology using highly realistic head models and outlines potential areas for future research. This review serves as a valuable reference for understanding blunt head injury mechanisms under ballistic impact helmet protection, optimizing target testing technology, and advancing related protection standards.

Keywords

Ballistic Impact, Helmet Protection, Blunt Head Injury Mechanisms, Target Testing Techniques, High-Simulation Head Model

1. Introduction

In high-risk scenarios such as warfare, military conflicts, and law enforcement operations, soldiers and police officers routinely face lethal threats from bullets and explosives. Ballistic helmets, as critical protective equipment, significantly enhance individual survivability across diverse environments by providing essential protection [1]-[6]. The protective mechanism of ballistic helmets relies on their ability to absorb and dissipate the kinetic energy of projectiles or fragments, thereby preventing penetration. However, despite successfully blocking penetrating injuries, high-velocity impacts may still cause deformation of the helmet's inner surface—known as Back Face Deformation (BFD) [1]-[6]. When such blunt trauma is transmitted to the head, it involves characteristic features including high velocity, substantial energy transfer, and short impulse duration. This impact first delivers a severe mechanical insult to the skull, potentially causing skull deformation or fracture [7]-[9]. Subsequently, forces propagate through the cranium to intracranial tissues, triggering complex biophysiological responses. These include brain tissue displacement, shearing, hemorrhage, and edema [10] [11], which can result in concussion, cerebral contusion, or even fatality. Consequently, wearers remain vulnerable to Behind Helmet Blunt Trauma (BHBT) even while protected by ballistic helmets. A comprehensive understanding of this injury mechanism is therefore crucial for advancing helmet protective performance and reducing operator morbidity and mortality.

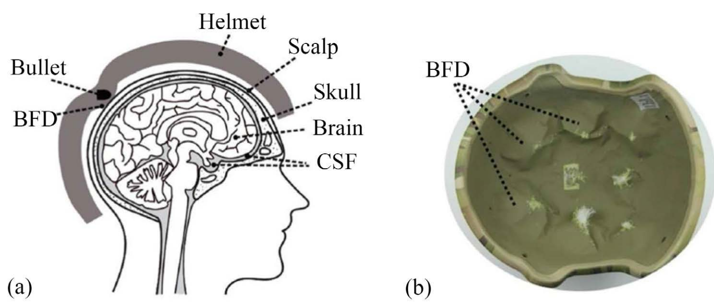

Figure 1. Deformation of the backside and blunt force trauma under ballistic impact helmet protection.

In order to comprehensively assess the protective effectiveness of ballistic helmets, national and international standards have been developed that are designed to ensure that helmets provide effective protection against different threats. In foreign countries, the NIJ 0106.01 standard is widely recognized and applied, which requires that helmets should not show penetrating and penetrating cracks when subjected to impact, and also requires that the peak acceleration be controlled within 400 g [12]. In addition, the U.S. Army standard further refines the requirements by stating that the BFD should be no more than 25.4 mm at the front and rear, and 16.0 mm at the left, right, and crown, which is achieved by measuring the depth of clay deformation and comparing it to thresholds at different impact sites to ensure that helmets are able to mitigate head injuries when subjected to nonpenetrating ballistic impacts [13]. In China, according to the standard GA 293-2012, bulletproof helmets must be successfully blocked in the case of 5 rounds of effective hits, and the height of the shell traces should be strictly controlled within 25 mm [14]. The effect of back and blunt force trauma deformation mechanism of lightweight helmets under ballistic impact is shown in **Figure 1**. The above criteria can be found to assess the protective performance of ballistic helmets mainly based on two key items, the helmet should stop the projectile and BFD, and the threshold of clay deformation has been used as one of the important indexes for assessing the protective performance of helmets, but in fact, the mechanism of correlation between the degree of clay deformation and the actual risk of injuries is not clear [15]. Therefore, in future research and standard-setting, there is a need to further explore and improve the assessment indicators and methods related to injury risk, so as to assess the protective performance of ballistic helmets with more accurate and reliable testing methods.

In order to assess the ballistic performance of ballistic helmets, helmet head moulds are usually mounted on a base and used in accordance with the standard, consisting of a rigid head and neck assembly and a padded witness material. In fact, bullet-helmet interaction involves three main dynamic events: firstly, the localised rear surface deformation (BFD) of the helmet due to bullet impact, a process that occurs in a very short period of time (<0.5 ms) and can lead to localised crushing of the helmet; secondly, there is the volumetric movement of the helmet head substance, a process that lasts for a long time (0.5 - 15 ms) and leads to acceleration of the helmet and the head; and lastly, there is the neck response induced by the impact of the bullet, a process that lasts for several milliseconds to tens of milliseconds (1 - 50 ms) [16]. The examination of each of these events is critical when assessing the ballistic performance of helmets, however, the head moulds currently in use are predominantly made up of rigid materials, which may differ from the dynamic response of the head and neck in real life situations. More importantly, it is unclear whether the perforation and BFD measures currently used to assess ballistic performance are sufficient to fully reflect the light and moderate injuries caused by BHBT. In order to more accurately assess the protective performance of ballistic helmets under the high velocity impact of bullets and fragments, future research should further explore and improve the testing methodology to more comprehensively consider the risk of BHBT.

Therefore, by reviewing the mechanism of blunt head injury under ballistic impact helmet protection, this paper provides an in-depth discussion of the damage that may occur to the head when it is subjected to a blunt ballistic impact, and also details the current evaluation criteria for blunt head injury. At the same time, the article also evaluates the advantages and disadvantages of these standards and their feasibility in the assessment of head injury from ballistic impact. Finally, it reviews the current target test methods for head injury assessment under ballistic impact, including Post-Mortem Human Subject (PMHS) targets, animal targets and simulated head mould targets. In particular, the latest progress in the frontier field of high-fidelity sensing head target is analysed in depth and outlooked, and the target test method not only provides us with a more accurate and reliable means of damage assessment, but also provides an important reference for the development and improvement of bullet-proof helmets.

2. Mechanism of Blunt Trauma from Ballistic Head Trauma

The mechanics of ballistic blunt impact is the basis of the injury causing process, when projectiles and fragments hit the helmet, they will first create a high pressure area at the point of impact, this pressure wave will rapidly propagate through the skull, the skull will try to disperse and absorb the energy when impacted, but the excessive pressure may lead to localised or overall deformation of the skull, or even fracture, meanwhile, the brain tissues will deform when impacted by the pressure wave, which may lead to neuronal damage, blood vessel rupture and nerve fibre breakage. Furthermore, in addition to direct pressure, the impact may generate shear and rotational forces, which are also capable of causing severe damage to brain tissue. Therefore, as there are multiple mechanisms by which a bullet impacts a helmet and acts on the head, the injury process is multifactorial, and this section focuses on the possible mechanisms of skull fracture and traumatic brain injury (TBI), as well as the biomechanical parameters and assessment metrics associated with these injuries.

2.1. Skull Injuries

The skull is an important organ for protecting human brain tissue, and when the head is subjected to blunt force loading, the skull is the first organ to sustain injury and the most serious damage, the rigid and elastic properties of the skull affect the distribution and attenuation of the impact force in the head, and when the deformation of the skull caused by cranial impact reaches the fracture tolerance of the skull, a fracture occurs, which further exacerbates the damage to the brain tissue, and based on the different characteristics of the impact, the resulting fractures are usually linear, depressed, rigid, round and multiple [17] [18]. The most common types of fracture under ballistic blunt impact are linear and depressed fractures, with the severity of injury increasing with increasing deformation of the back of the helmet, from simple linear fractures to a combination of linear and depressed fractures, with many factors influencing the cranial response including impact velocity, contact area and loading area, with impacts with a wide contact area (typically greater than 13 cm^2) tending to cause linear fractures [19], while relatively localised impacts tend to result in depressed fractures [8]. A research by Rafaels *et al.* [20] showed that BFD induced by high velocity (460 m/s) bullet impact produced only linear fractures, whereas moderate velocity (440 m/s) impact resulted in linear and depressed fractures. In addition, the location of the bullet impact affects fracture formation, and fracture tolerance varies in different regions of the human skull; for example, the forehead can withstand greater impact forces than the zygomatic bone or the mandible [7] [9], and Sarron *et al.* [21] reported that when the distance between the skull and the protective polyethylene sheet was increased from 12 to 15 mm, the mean linear fracture length decreased from 15.3 cm to 6.6 cm, based on these insights into skull fracture, the design of protective helmets needs to take into account the helmet material, the distance between the helmet and the head, and the protective properties of different areas of the helmet.

2.2. Traumatic Brain Injury

The complex and multifactorial nature of traumatic brain injury, more so than skull fracture and not yet fully characterised, is reflected in the diversity of its aetiology and mechanisms. Mechanical loads such as skull deformation, head translation acceleration, rotational acceleration and rotational velocities can lead to complex strains in brain tissue, including shear, stretching and compression, which can lead to a range of potential brain damage, which is the same as that caused by ballistic blunt force trauma The potential mechanisms of ballistic blunt force. Previous studies have attempted to correlate intracranial parameters of the head, such as intracranial pressure (ICP), with brain injury [11], and the intracranial response, which may originate from one or more of these mechanisms, is important for understanding and assessing TBI. TBI can be categorised into focal brain injury and diffuse brain injury. Focal brain injury usually refers to an injury caused by an external force acting directly on a specific area of the head, and this type of injury often results in the death or loss of function of nerve cells in that area. Blunt ballistic impacts can lead to focal brain injuries occurring below the impact site. Skull deformation and relative cranio-cerebral motion (caused by acceleration) are potential causes of focal brain injuries. Skull deformation transmits compressive stresses to the brain tissues below, leading to focal brain injuries such as cerebral contusions and haemorrhages. When skull deformation is fractured, the meningeal blood vessels may rupture and lead to an epidural haematoma [19]. In addition, the skull may move relative to the brain after impact due to inertial effects, and this relative linear motion can result in the inner contours of the skull colliding with the brain, leading to focal injuries.

Diffuse brain injury, on the other hand, is a condition in which multiple regions of the brain are damaged simultaneously due to an external force. Angular and linear acceleration have long been associated with diffuse axonal injuries (DAIs) and cerebral haemorrhage in blunt impacts [10] [22]. Studies of blunt impacts in general have shown that, due to the local change in density, neighbouring parts undergo different inertial motions and angular acceleration induces shear stresses in the tissue. Meanwhile, translational acceleration induced by blunt impact can lead to concentration of shear stress on the corpus callosum and brainstem leading to DAIs [12], DAIs were reported in a study by Anderson *et al.* [23], which used a lightweight projectile to impact anaesthetised sheep, which found a positive correlation between linear acceleration and angular acceleration and the severity of DAIs. In conclusion, due to the lack of experimental studies and detailed clinical data, the main causes of diffuse traumatic brain injury are unknown and require further scientific endeavours.

3. Evaluation Criteria for Blunt Head Injury

3.1. Head Injury Criteria (HIC)

Currently, the most widely accepted method for assessing head injury risk in road safety research is the Head Injury Criterion HIC (HeadInjuryCriterion), which is used to assess the risk of brain injury in vehicle occupants, based on the criterion of linear acceleration at the center of gravity of the head [4] [24], which is a weighted product of acceleration and impact duration and has been found to correlate with the severity of brain injury, and can be used as a measure of the likelihood of a head injury resulting from an impact [2]. Commonly used to assess the risk of occupant head injuries in vehicle crashes, as well as the safety of vehicles, personal protective equipment and sports equipment, etc.

$$\mathrm{HIC} = \max\left\{\left(t_2 - t_1\right)\left[\frac{1}{t_2 - t_1}\int_{t_2}^{t_1} a(t)\,dt\right]^{\alpha}\right\} \quad (1)$$

where a(t) is the linear acceleration time scale as a multiple of g, and $\alpha = 2.5$, are the two multiplier moments chosen in a way that maximises the HIC result, subject to the constraint that they do not differ by more than a specified interval. The impact time intervals of 15 ms and 36 ms corresponding to HIC15 and HIC36, respectively, have been widely used for automotive safety applications. The higher the HIC value, the higher the risk of serious injury in order to analyse shorter duration impacts and determine the HIC value, an injury assessment can be made by referring to the Abbreviated Injury Scale (AIS), which estimates the probability of a specific degree of craniocerebral injury for a specific HIC value [25].

The HIC criterion has also been used in ballistic studies to prevent head injuries, e.g. Yang *et al.* [26] used HIC to study the endpoint effect of a ballistic helmet head model at different impact angles and at different impact locations but the results were inconclusive as it was not related to any injury mechanism. Since HIC only considers the risk of injury to the head due to linear acceleration of the head and cannot be used to capture local deformation caused by blunt ballistic impact and predict the likelihood of head injury due to head rotation, nor can it distinguish between different types of head injuries that may occur, such as skull fractures, subdural haematomas or diffuse axonal injuries [27], the HIC criterion is unsuitable for the study of blunt impact injuries of the head due to ballistic impact [26]. studies, and the unique impact loads of blunt ballistics require new injury assessment criteria and methods [27] [28].

3.2. Blunt Collimation (BC)

The bluntness criterion is commonly used as a predictor of skull fracture because it takes into account both impactor and head parameters, and the BC has a higher chance of correct prediction compared to other parameters [29], mainly because it captures specimen soft tissue thickness, which is an important determinant of skull fracture [29], and the BC is also able to capture the effects of impactor characteristics (e.g., mass, contact area, and kinetic energy), which have proven to be particularly important when assessing skull fracture contact area has been shown to be particularly important when assessing risk of skull fracture.

$$BC = \ln\left(\frac{mV^2}{2M^{\frac{1}{3}}TD}\right) \tag{2}$$

where $mV^2/2$ is the impact energy in J, m is the mass of the head in kg, T is the thickness of the skull in cm, and D is the diameter of the impactor in cm. In a study by Raymond *et al.* [30] a BC of 1.61 corresponded to a 50% risk of skull fracture, and Bailey *et al.* [31] carried out extensive tests using a bovine scapula model to further investigate skull fracture criteria for blunt ballistic impacts, the BC threshold for 50% fracture risk was determined to be 1.1 - 1.3, the impact surface was flat, the BC was sensitive to the shape of the impactor surface, the thresholds for flat projectiles were significantly different from those for curved projectiles, with further refinement, e.g. by incorporating the shape of the impactor, the BC may be a suitable predictor of skull injury due to blunt ballistic impacts [19].

3.3. Criteria for Assessing Intracranial Pressure (ICP)

Intracranial pressure (ICP) is the pressure of cerebrospinal fluid in the cranial cavity, which normally remains stable. However, when a blunt head injury occurs, it may result in brain tissue damage, haemorrhage or oedema, which in turn causes an increase in intracranial pressure. ICP is one of the criteria for traumatic brain injury. Due to the observation of rapid changes in intracranial pressure during ballistic blunt impact, many BBHI studies have attempted to link brain injury to changes in intracranial pressure, cerebrospinal fluid pressure and brain parenchyma [32] [33]. Sarron *et al.* [21] in their protected human cadaver head impact tests found a correlation between ICP in a large pool and the occurrence of brain injury, with an increase in impact leading to an increase in ICP, which was shown to occur at ICP values greater than 191 kPa for severe cranial injuries. It is noteworthy that the type and severity of brain injury was not reported. Freitas *et al.* [5] reported an ICP range of 90 kPa to 255 kPa, which is a 9mm bullet impacting a head target protected by a USMC lightweight helmet with an impact velocity range of 428 - 480 m/s. The ICP range of 90 kPa to 255 kPa is the range of impact velocities for a 9 mm bullet impacting the head target. Anderson *et al.* [23] found a positive correlation between peak impact force, peak kinematics, peak ICP and severity of DAI using a sheep animal model, although the results obtained using an animal model may not be directly applicable to humans. Liu *et al.* [34] investigated the changes in brain parenchymal pressure (ICP) induced by non-penetrating 9 mm bullets in anaesthetised pigs. This study reported increases in maximum ICP with mean values of 219 kPa, 474 kPa, and 751 kPa, corresponding to impact velocities of approximately 280 m/s, 360 m/s, and 428 m/s, respectively, followed by peak pressure times of 0.311 ms, 0.238 ms, and 0.181 ms, respectively.

Despite the above studies, the threshold of ICP associated with BBHI brain injury has not been determined. A non-lethal ballistic impact study by Oukara *et al.* [35] obtained a craniocerebral injury threshold for ICP. A maximum ICP of 25 kPa - 45 kPa was found to correspond to a lack of consciousness in the affected specimen, a threshold that may be directly applicable only to the projectile used in this study, given the possible influence of the shape and quality of the firing pin on the mechanical response of the head and the outcome of the injury. Based on ICP thresholds established from numerical simulations of general blunt impacts in animal and human cadaveric models, ICP exceeding 173 kPa can lead to mild brain injury and ICP exceeding 235 kPa can lead to severe brain injury [36]. Studies have shown that the head can tolerate higher ICPs for shorter periods of time; therefore, the blunt impact thresholds described may not be directly applicable to blunt ballistic impacts with higher impact velocities and shorter ICP durations, and further study of ICP in BBHIs is necessary to improve our understanding of the injury mechanisms and to provide better criteria for injury assessment.

4. Target Testing Techniques

The significance of BBHI blunt trauma assessment is to deeply understand and predict the degree of injury that the head may suffer from ballistic blunt impact, this assessment not only focuses on the protective effect of the helmet itself on the impact, but also involves the biomechanical response of the head tissues after the impact and the possible damage, in order to carry out an effective assessment, testing is one of the key links, the target test can help the researchers to more deeply understand the targeting test can help researchers to better understand the biomechanical response of the head after an impact.

4.1. PHMS Target Testing

The PMHS is biologically closest to a living organism becoming a common test target. Weisenbach *et al.* [16] conducted a ballistic impact helmet protection head blunt force test using three cases of PMHS and showed that the mean impact force of impacts leading to skull fractures was 7,995 N (5,129 N to 10,883 N), however, this study did not correlate the impact force with the risk of injury. Rafaels *et al.* [20] used experimental cadaveric tissue to explore head injuries caused by BABT, using seven PMHS head/neck specimens wearing ballistic helmets exposed to non-perforating impacts at velocities ranging from 400 - 460 m/s, with an increasing trend in injury severity, from simple linear fractures to a combination of linear and depressed fractures. Sarron *et al.* [21] used dried skulls filled with silicone gel and fresh cadaveric heads to study the effects of helmet BFD, and the dried skulls resulted in much higher intracranial pressures due to the absence of the scalp and bone dehydration process compared to fresh cadaveric heads. Postmortem human subjects provided good anatomical performance; however, they were unable to mimic the physiological responses of living cells and tissues associated with respiratory and circulatory activity. As a result, PMHS is unable to realistically replicate responses related to circulating pressure and haemorrhage and does not allow for transducer embedding, which may affect the realism of mechanical responses (e.g., intracranial pressure and brain displacement) and injury responses (e.g., intracranial haemorrhage) Despite these concerns, PMHS is commonly used in injury studies, particularly in skull fracture studies, as the degree of alteration of skeletal responses is less than that of soft tissues Smaller.

4.2. Animal Target Testing

An alternative is provided by animal models, which are physiologically similar to humans and are often more readily available than PMHS, and can therefore mimic to some extent the reactions that may occur in humans in response to injury. Animal models used in impact studies include monkeys, pigs, dogs, sheep and calves [20] [37] [38]. Live animals provide physiopathologic models to study brain pathology, which is not possible with PMHS and cephalic models [37]. For example, through experiments with live pigs, Liu *et al.* [34] were able to measure ICP without altering the physiological and mechanical behavior of living tissues, and they concluded that stress in the brain is a reliable predictor of BBHIs. But the same unavoidable differences and ethical issues exist with human subjects.

Isolated tissues from animals have also been used for BBHIs studies. For example, the bovine scapula has been proposed as a replacement for the human cranium due to its similar thickness and geometry. Bailey *et al.* [39] improved the bovine scapula model by adding two layers of chamois leather to represent the scalp. Bovine scapulae were readily available, making large-scale experiments possible. In the context of brain injury, isolated animal brains, including bovine and rat, have been used to study pressure properties and material properties of the brain [38] [40]. The vitro models can facilitate understanding of material biomechanics; however, vitro animal models are not used for helmet performance testing due to differences in geometry and the inability to study brain-skull interactions with these models. Concerns about reproducibility and ethics also limit the use of vitro models.

In order to more accurately understand and predict the biomechanical response of humans under specific conditions, researchers have conducted dimensional analyses based on scenarios such as blunt-head impacts and explosions in an attempt to scale biomechanical responses and kinematic features, and these studies have involved the scaling of kinetic responses such as accelerations, forces, moments, and shockwaves, which were calculated based on the ratio of dimensions, densities, and moduli between species [41]-[43]. However, it is worth noting that these scaling laws fall short in terms of validation against biomechanical head injury (BBHI) loading scenarios, where validity is often influenced by mechanical loading conditions that determine head injury mechanisms and biophysical responses. For example, brain size does not have a significant effect on brain pressure in blast scenarios, but plays an important role in blunt impact [42]. This means that brain pressure does not need to be scaled between species in blast scenarios, but needs to be scaled accordingly in blunt impacts. Thus, the direct application of results from animal studies to human studies faces significant challenges due to the significant morphology and differences between species.

4.3. Head Mold Target Test

Testing of PHMS targets as well as animal targets is very complex due to technical, ethical and legal reasons. Therefore, it is important to develop alternative simulated head targets and corresponding test methods in order to avoid testing PMHS and animals. The head model is durable, reproducible and easy to use.

4.3.1. Hybrid III Dummy Head Molds

The Hybrid III dummy head model was originally developed for automotive crashes, integrated with an accelerometer configuration in the center of the head. The Hybrid III head has been used to study head acceleration from ballistic impacts on combat helmets [44]. The Hybrid III head has also been modified to include surface pressure sensors at various points in the head mold to measure impact pressure; however, impact forces or pressures measured using the Hybrid III head may not be the same as those sustained by the human head due to differences in the hardness of the scalp and skull. Compared to the PMHS, the Hybrid III head consists of a softer vinyl scalp and a stiffer aluminum skull, and Raymond *et al.* [12] found that the impact force of the Hybrid III head mold was significantly greater than that using the PMHS in ballistic blunt impact.

4.3.2. BLSH Head Molds

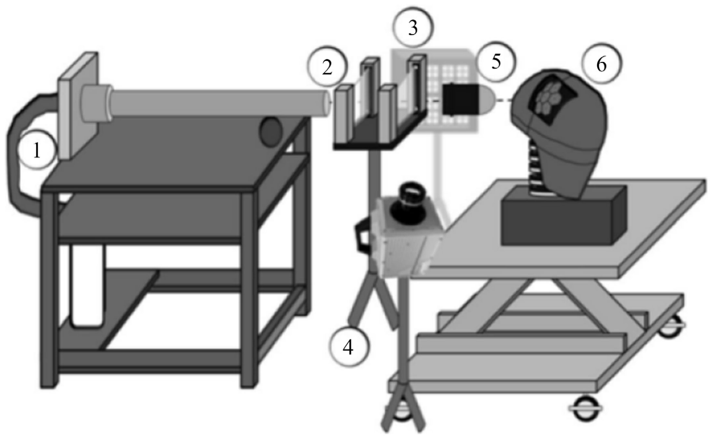

Figure 2. BLSH head mold test system [35].

The BLSH head mold test system is shown in **Figure 2**. The BLSH head mold was developed by Biokinetics and Associates to directly measure the dynamic loads applied to the skull as a result of a non-penetrating projectile impact. This BLSH was initially developed to assess the risk of skull fracture due to posterior facial deformation of military helmets upon impact with bullets [35]. Raymond *et al.* [45] compared the forces measured on the PMHS with those measured using the

BLSH and other alternatives and concluded that the BLSH was the most appropriate for assessing the forces of head impacts of non-lethal projectiles. The BLSH is equipped with seven Kistler impact force sensors to measure contact force. The cranial substructure is made of magnesium and the silicone rubber pads are used as skin substitutes covering the load cell arrays including four impact positions: lateral (left, right), anterior, and posterior, but the BLSH limitation is that it has sensors only on the frontal or lateral sides of the head mold, material simulation materials, including skin, bone, and soft tissues, etc. on.

4.3.3. High-Simulation Head Molds

Hammouk *et al.* [16] conducted ballistic experiments on three different head models, and the structure of the specific simulation experiment is shown in **Figure 3**. a rigid head model, a Hybrid-III based flexible head model, and a high simulation based head model, and investigated the response of the head agent in the anterior, posterior, lateral, and coronal directions, and measured the posterior facial deformation (BFD) of the head morphology, the head kinematics, and the intracranial pressure, and the results showed that the type of the head model influenced the biomechanical response, with a statistically significant increase in head kinematics for the flexible head model compared to the rigid head model.

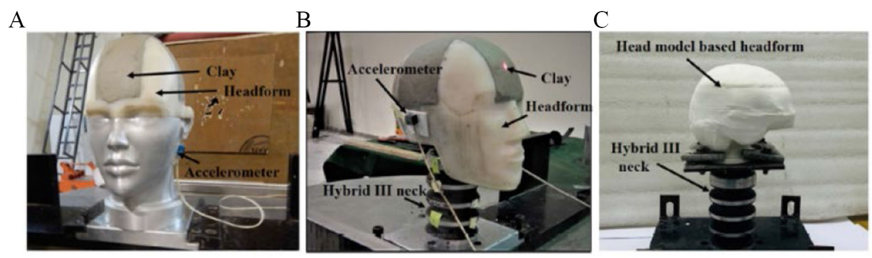

Figure 3. Harmukh *et al.* study on high simulation head molds [16].

Freitas *et al.* [5]measured the dynamic response induced by blunt trauma at the back of a helmet using a human head substitute, a unique human head substitute based on a renewed human skull and alternative materials representing the soft tissues of the human head (e.g., skin, dura mater, and brain), and sensors embedded in the human head substitute that allow for direct measurements of intracranial pressure, intracranial pressure, and head and helmet acceleration. Merkle *et al.* [46] investigated the brain's response to excitotube overpressure loading conditions using a high simulation head model consisting of the brain, skull, facial structures, and skin, all of which were fabricated using biomimetic materials, which was mounted on the neck of the Hybrid III anthropomorphic test device to allow for head movement during overpressure exposure. The high simulation head model used for intracranial pressure testing is shown in **Figure 4**. The pressure transducers were embedded along the sagittal plane in the anterior and posterior regions of the biomimetic brain, and a series of excitotube tests were conducted using four levels of actuator pressure (ranging from 420 to 1,150 kPa) to simulate blast loading conditions consisting of head structures made of biomimetic materials, such as a glass/epoxy resin blend for the skull, and Sylgard silicone gel for the brain.

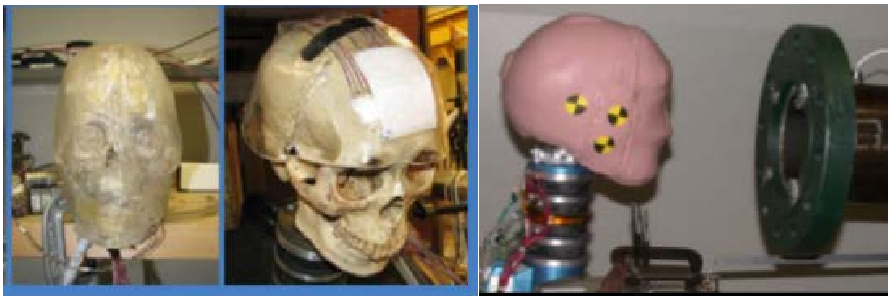

Figure 4. High-simulation head model for intracranial pressure testing [5] [46].

Obviously, in the future, in high-precision biomechanical modeling, advanced material technology, high-precision sensing technology, integrating a variety of high-precision sensors (e.g., pressure sensors, acceleration sensors, displacement sensors, etc.) into a high simulation head model, in order to real-time capture and record a variety of physical parameters in the process of impact or collision. Develop multi-parameter fusion algorithms for comprehensive analysis and processing of sensor data to more accurately assess the degree and type of head injuries, promote the standardization process of high-fidelity head model technology, formulate unified testing methods and assessment standards, and develop simulation materials that are closer to real human tissues, including skin, bones and soft tissues, in order to improve the accuracy and reliability of testing. Integrate multiple sensors (e.g., pressure, acceleration, displacement, etc.) into the simulated head model to realize multi-parameter and multi-dimensional data acquisition and analysis. In-depth study of the biomechanical properties of the human head, including the deformation, stress distribution and damage mechanism of the head under impact, etc., in order to optimize the design and testing methods of the simulated head model.

5. Conclusions

In order to deeply understand and evaluate the protective effect of helmets on the head under ballistic impact and the mechanism of blunt trauma it induces, this paper reviews the research progress on the mechanism of blunt head trauma under ballistic impact helmet protection and target testing:

1) The biomechanical response is very complex and involves the interaction of multiple components of the skull, brain tissue, cerebrospinal fluid, etc. Multiple mechanisms occur during the blunt impact of the bullet hitting the helmet and acting on the head, which are mainly involved in typical blunt injuries such as skull fractures and traumatic brain injury.

2) The assessment of the protective performance of ballistic helmets at home and abroad is mainly based on two key items, the helmet should stop the projectile

and BFD, while the threshold of clay deformation is used as one of the important indicators for assessing the protective performance of helmets, and it is not clear whether the measures of perforation and BFD, which are used for assessing the ballistic performance, are sufficient to fully reflect the light and moderate injuries caused by BHBT, so we need to further explore and improve the assessment methodology in order to more accurately assess the protective effect of helmets.

3) Blunt guidelines may be a suitable predictor of cranial injury due to blunt ballistic impact; however, it remains a challenge to accurately measure this metric. In addition, intracranial pressure as a more promising intracranial measure for predicting the risk of traumatic brain injury in patients with ballistic brain injury deserves further research and attention.

4) The future development of high simulation head model technology will pay more attention to high precision, multi-parameter fusion and standardization, as well as closer to the real human body tissue simulation material development, which will provide strong support for helmet design and safety standards.

Conflicts of Interest

The authors declare no conflicts of interest regarding the publication of this paper.

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