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Behind helmet blunt trauma induced by ballistic impact: A computational model

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ABSTRACT

Behind helmet blunt trauma (BHBT) has emerged as a serious injury type experienced by soldiers in battlefields. BHBT has been found to range from skin lacerations to brain damage and extensive skull fracture. It has been believed that such injuries are caused by forces transmitted from the helmet's back face deformation (BFD), which result in local deformations of the skull and translation or rotation of the head, leading to brain injuries. In this study, head injury risks resulting from the BFD of the Advanced Combat Helmet (ACH) under ballistic impact are evaluated using finite element simulations. The head model developed at KTH in Sweden is adopted, and a helmet shell model (including foam pads) is constructed. The examined mechanical parameters include the maximum von Mises stress in the skull, pressure (mean normal stress) and maximum principal strain in the brain tissue, contact force, and head acceleration. The influences of the foam pad hardness, stand-off distance, helmet shell thickness, and impact direction on head injury risks are studied. It is found that a softer foam pad offers a better protection, but the foam pad cannot be too soft. Also, it is shown that a slightly larger stand-off distance leads to a significant reduction in head injury. In addition, the simulation results reveal that an increase in the helmet thickness reduces the injury risk. It is further observed that a 45-degree oblique frontal impact results in a lower head injury risk than a 90-degree frontal impact. Moreover, for a helmet protected head under ballistic impact, it is seen that a high risk of skull fracture does not necessarily mean an equally high risk of injury to the brain tissue. The predictions from the current model of a helmeted head under ballistic impact agree with experimental findings independently obtained by others. The newly developed model provides a useful tool for studying injury mechanisms of BHBT and evaluating the existing standards for testing and designing combat helmets.

1. Introduction

Modern combat helmets made from advanced composites provide good protection against penetrating head injuries from ballistic and shrapnel threats and have saved lives of many soldiers (e.g., Kulkarni et al., 2013 [1]; Jenson and Unnikrishnan, 2015 [2]). Currently serving combat helmets are designed to be lighter than older ones (e.g., Hisley et al., 2011 [3]; Kulkarni et al., 2013 [1]; Freitas et al., 2014 [4,5]). The reduction in the helmet weight tends to result in a larger back face deformation (BFD), which can lead to head injuries known as behind helmet blunt trauma (BHBT) for soldiers in battlefields [5–7]. BHBT has emerged as a serious injury type experienced by soldiers (e.g., Carroll and Soderstrom, 1978 [8]; Sarron et al., 2000 [9]; Cannon, 2001 [10]; Hisley et al., 2011 [3]; Prat et al., 2012 [11]). BHBT has been found to range from skin lacerations to brain damage and extensive skull fracture (e.g., Freitas et al., 2014 [5]). Such injuries are caused by forces transmitted from the helmet's BFD, which result in local deformations of the skull and translation or rotation of the head, leading to brain injuries [6]. BFD is a basic measure for the ballistic performance of combat helmets in the current testing standards. However, injury mechanisms associated with BHBT arising from helmet BFD are still poorly understood (e.g., Young et al., 2015 [12]).

A few experimental studies have been conducted to gain insights into head injuries induced by non-penetrating impacts. An early investigation by Sarron et al. (2000) [9] utilized dry human skulls filled with silicone gels, which were impacted under protection of aluminum plates. The contact force and pressure on the skull surface were recorded. A further study conducted by the same group [13] using human cadaver heads protected by aluminum and composite plates revealed injuries ranging from skin lacerations to extensive skull fracture. In particular, the intracranial pressure (ICP) was quantified in Sarron et al. (2004) [13], which provides valuable information about brain tissue injury. However, these studies

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are of limited values, since a flat plate (rather than a real helmet) positioned a few millimeters from the head was used. In a more recent study by Rafaels et al. (2015) [7], non-perforating impacts were conducted on postmortem human subject (PMHS) head/neck specimens wearing military protective helmets, and a detailed examination of the resulting injuries, especially skull fracture patterns, was performed qualitatively. Non-penetrating ballistic impacts on live, anesthetized pigs were performed by Liu et al. (2012) [14], in which the physiopathological changes in living tissues were included and the ICP was analyzed. To better characterize head injuries associated with helmet BFD, a human head surrogate was recently developed by Freitas et al. (2014) [5] using refreshed human craniums and surrogate materials that represent human head soft tissues (such as the skin, dura, and brain). The extensive information obtained in Freitas et al. (2014) [5], including measured ICP, cranial bone strain and head acceleration, provides new insights into head injuries resulting from non-penetrating ballistic impacts on combat helmets.

Testing of a helmeted head to study injuries of the skull and brain tissue has been a great challenge. Variants of head surrogates and helmets from different manufacturers make comparisons difficult. Furthermore, specimens are difficult to obtain, and tests can be lengthy and expensive. In view of these, finite element (FE) models have been employed to simulate performance of various helmets. The findings based on such models can provide valuable guidance for evaluating and designing helmets to attenuate head injuries.

By using a validated head model including detailed anatomical features, Aare and Kleiven (2007) [15] found that the helmet shell stiffness should neither be too stiff nor too soft to achieve a good protection. Additionally, they showed that an oblique ballistic impact may cause more injuries in the brain tissue than a pure radial one. A further study on different interior systems revealed that a low frictional layer can be added to reduce the brain injury risk under ballistic impact [16,17]. Tan et al. (2012) [18] conducted both experimental and numerical studies to investigate the performance of the Advanced Combat Helmet (ACH) under ballistic impact. While a good correlation was found for head acceleration, helmet deflection and helmet damage, they were unable to evaluate local brain injury because of the use of a simple dummy head model. FE models of a human head including its principal anatomical features have been used in the ballistic study of Baumgartner and Willinger (2005) [19] by simplifying a helmet as an aluminum plate. The protective role of a Personnel Armor System Ground Troops (PASGT) helmet with strap nets was simulated by Lee and Gong (2010) [20] using a head model, which indicated that both the intracranial brain pressure and head acceleration exceeded their respective threshold values for serious brain injuries. A more recent study by Tse et al. (2014) [21] examined the difference between the strap-netting and Oregon aero foam padding. In another study by Jazi et al. (2014) [22], the influence of foam pad stiffness on head injury was evaluated. As reported in Tse et al. (2014) [21], the maximum helmet deflection was around 10.9 mm, while less than 12 mm was reported in Jazi et al. (2014) [22]. Considering that experimentally obtained helmet deflection values are normally larger than 25 mm in helmet shell testing [3,6], a major limitation of the above-mentioned models is that the loading conditions are not representative of actual ballistic impact events. The models that did consider realistic ballistic loading conditions examined only the maximum value of BFD. But dynamic deflections of the helmet shell, which may have a large influence on the blunt impact to the head, have not been studied.

The mechanisms of head injuries associated with helmet BFD remain unclear, although considerable research efforts have been made. According to the current testing protocols for combat helmets, the resistance to penetration (RTP) and BFD are the two primary measures. This, however, has been found to be questionable [6]. In particular, whether the BFD is appropriate for assessing how well a helmet protects the head has become a major concern, since there is no scientific basis supporting a correlation between the BFD and head injury [6]. A further insight into the injury mechanisms is therefore critical not only to address the urgent need for developing better helmet test metrics but also to help guide future combat helmet designs for a better protection. This motivated the current work.

The present paper aims to develop a computational model to study ballistic performance of combat helmets and to investigate head injury mechanisms associated with the helmet BFD induced by non-penetrating ballistic impact. A new finite element model for a combat helmet with foam pads is developed and fitted onto a head model constructed earlier. The helmet model is validated against existing experimental data for the maximum value and time history of helmet BFD. In addition, the effects of foam pad hardness, stand-off distance, helmet shell thickness, impact direction, and on- and off-pad impacts on head injury risks are studied.

2. Model description

The finite element models for the helmet shell, bullet, foam pads and head are described first, which are followed by the assembly of the helmet and the head model. All of the computational simulations are performed using the finite element code LS-DYNA (2015) [23].

2.1. Helmet

2.1.1. Helmet shell modeling

The Advanced Combat Helmet (ACH) is considered in this study. ACH is the currently serving combat helmet in the U.S. Army and is made of layers of Kevlar 129 fibers bonded by a thermoset resin as the matrix material (e.g., Kulkarni et al., 2013 [1]).

The geometrical model of an ACH is meshed using hexahedron solid elements. The mesh density is refined at the sites of impact (see Fig. 1(a)). The convergence study has revealed that the mesh used here is adequate in obtaining a converged numerical solution [24]. The helmet shell thickness in the model is 7.8 mm [25]. The foam pads used in the ACH, which consist of two layers with a soft layer close to the head and a hard layer attached to the helmet shell, are manufactured from Team Wendy polyurethane foams [26,27]. The foam pad configuration shown in Fig. 1(b) includes seven pads located at the front center, front right, front left, back right, back left, back center, and on the crown.

In this study, the woven fabric-reinforced laminated composite used to make the helmet shell is regarded as an orthotropic material, which is represented using nine elastic constants including three Young's moduli E_{11}, E_{22}, E_{33}, three Poisson's ratios ν_{12}, ν_{13},

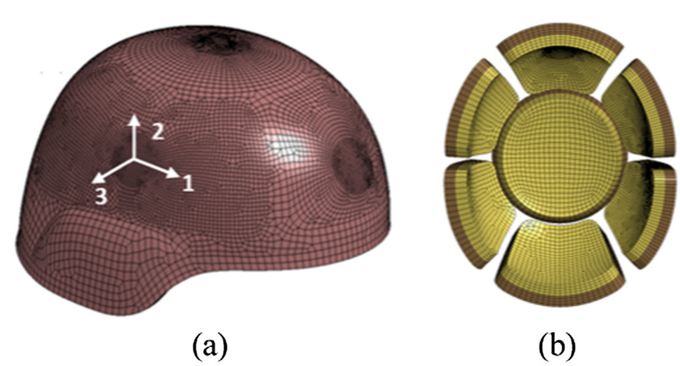

Fig. 1. (a) FE mesh of a large-size ACH shell and (b) FE mesh of foam pads. Here, "1" and "2" represent the two in-plane directions and "3" stands for the thickness direction.