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A COMPUTATIONAL MODEL FOR OPTIMIZATION DESIGN OF CONSTRUCTION HELMET

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SUMMARY

Construction helmets are essential personal protective equipment for reducing exposure to traumatic brain injury at work sites. We proposed a finite element modeling approach that would be suitable for engineers to optimize construction helmet design. The model includes essential anatomical structures of a human head (i.e., skin, scalp, skull, cerebrospinal fluid, brain, medulla, spinal cord, cervical vertebrae and discs) and major engineering components of a construction helmet (i.e., shell and suspension system). We demonstrated the evaluation of the performance of a construction helmet using the proposed computational method.

Key words: *finite element analysis, brain injury, helmet design*

1 INTRODUCTION

Traumatic brain injuries (TBIs) are among the most common severely disabling injuries in the United States; during 2002-2006, approximately 1.7 million cases occurred in civilians annually [1]. Work-related TBIs occur frequently in such industries as construction. A total of 7294 work-related TBI fatalities were identified during 2003-2008, which accounted for 22% of all occupational injury fatalities [2]. Among the leading causes of work-related TBI death, falls and contact with objects/equipment represented 47% [2]. The work-related TBI fatalities due to contact with objects may be reduced by using properly designed and manufactured helmets.

Finite element (FE) models have not only been used in the investigation of injury mechanisms [3], but also in the design of head protective systems [9]. For example, Afshari and Rajaari [5] developed FE models to study the protective effectiveness of the helmet during the head-ground impact of a motorcyclist. Teng *et al.* [6] developed FE models of a bicycle helmet with foam liners and validated their model with impact tests. Although these models included detailed helmet geometries and material properties, they did not include realistic anatomical structures of the human head. Yang and Dai [7] developed FE models to study the ballistic helmet impact; their models included realistic geometries and material properties of the helmet and human head. These models have been further developed by Long *et al.* [8] to assess the performance of construction helmets.

Most of the previous head-brain models are used for frontal impacts and do not include the neck. It is widely believed that the effects of the neck and body mass on the brain responses during short impact intervals (duration less than 7 milliseconds) are negligible [4]; however, the effects of the neck and body mass have not been quantified. Our goal is to develop a practical FE model that would include essential anatomical details of the human head-brain; at same time, it would be small enough to be suitable for engineers to optimize construction helmet design.

2 METHODOLOGY

2.1 Finite element model

The helmet model consisted of a shell and a suspension system (Fig. 1A-B). The shell geometry was obtained by scanning a representative, commercially available construction helmet (Model V-Gard, MSA Safety Inc., Pittsburgh, PA, USA). The geometry of the suspension system was constructed using commercially available software Solidworks (Autodesk, Inc., San Rafael, CA, USA). The 3D geometries of the shell and suspension were then imported into ABAQUS (Abaqus US/Feasol, Boston, MA) to generate FE meshes. The model of the helmet shell was constructed using shell elements, whereas that of the suspension system was generated using 3D continuous elements. The suspension system was constrained to the helmet shell at four plug locations.

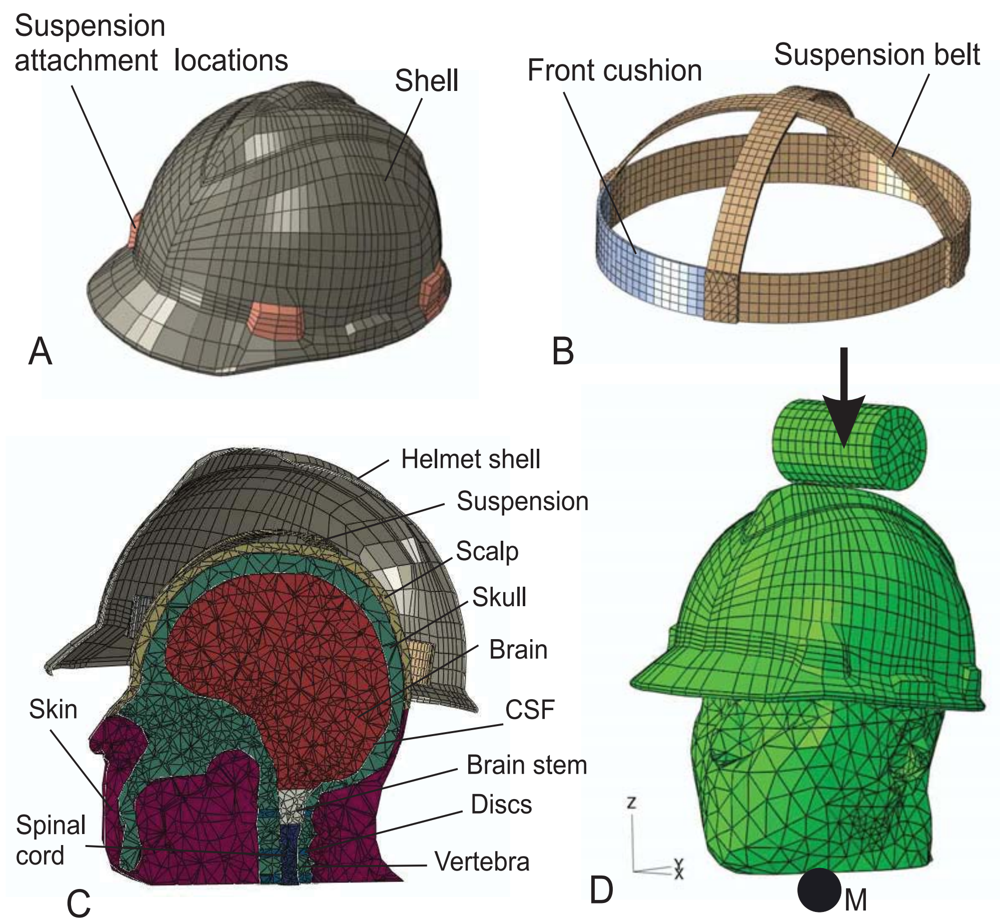

Figure 1: FE model of the head-helmet. A: Helmet shell. B: Helmet suspension system. C: Cross-sectional view of the head-helmet complex. D: Simulation of a top impact.

The FE meshes of the head-brain-neck complex were developed by using a commercially available data base (Materialise, Leuven, Belgium). The dimensions of these head surface meshes represent approximately the 50th percentile of Caucasian males. The head-brain-neck complex consisted of scalp, skin tissues, skull, cervical vertebrae (C1, C2, and C3), discs, brain, medulla, cerebrospinal fluid (CSF), and spinal cord (Fig. 1C). The brain tissues included the cerebrum, cerebellum, and a part of the brain stem (midbrain and pons) (Fig. 1C). The spinal cord included the surrounding *pia mater*. The CSF was considered to cover the entire external surface of the brain, medulla, and the spinal cord. The discs contained both *annulus fibrosus* and *nucleus pulposus*. Within each of these components (i.e., brain, medulla, CSF, spinal cord, and discs), the material was considered homogeneous. The connections between the tissues were assumed perfect bond, without relative sliding during deformation. The CSF (thickness 1.3 mm) was constructed using membrane elements, whereas all other components were constructed using three-dimensional continuous elements. The falling object was cylindrical (diameter 28.5 mm, length 100 mm) and was modeled using 3D continuous elements (Fig. 1D). A point mass of 10 kg was connected to the vertebral bone at the neck, simulating the inertial effects of the rest of the body during impact.

2.2 Material properties

The helmet shell was considered to be made of typical ABS plastic. The suspension top belt side ring was considered to be of high strength polymers. The front cushion of the suspension system was of soft foam material. The falling cylinder was considered to be made of steel and had a mass of 2 kg. All materials of the helmet components were considered to be linearly elastic. The scalp, skull bone, cervical discs, and vertebral bone were considered to be linearly elastic. The CSF was considered as a weak, elastic and nearly incompressible medium. The skin, brain, medulla, and spinal cord were considered to be hyperelastic and viscoelastic. The finite deformation formulation was used in describing the constitutive models due to large tissue deformations.

2.3 Simulations

Using the proposed model, we simulated the impact force and brain acceleration during an impact of an object on top of the helmet. Initially, the cylinder was at a height of 3.27 m above the helmet top and it fell due to gravity; it reached a speed of 8 m/s just before impacting on the helmet (Fig. 1D).

3 RESULTS AND CONCLUSIONS

The impact force between the cylinder and helmet shell as well as that between the head and suspension system as a function of the time are shown in Fig. 2A. The simulations indicate that the cylinder bounced and separated from the helmet shell (impact force became zero) after the initial impact; and that the impact force between the head and suspension system did not reach the peak at the same time as that between the cylinder and helmet shell. The ratio of the initial peak of F_b (contact force of head-helmet suspension) to that of F_a (contact force of object-helmet) is ap-

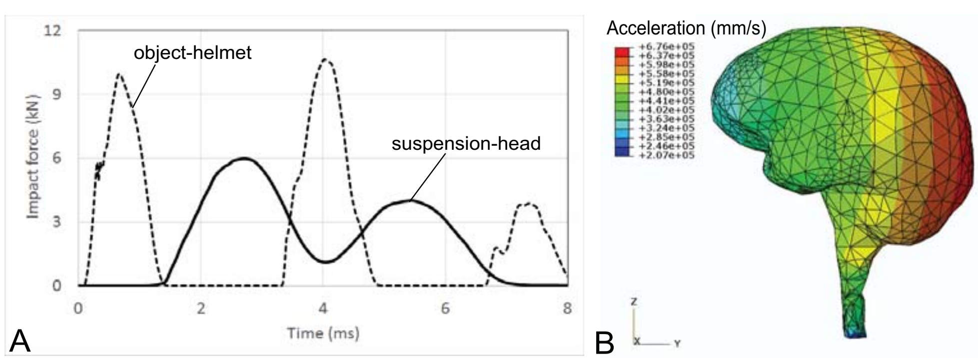

Figure 2: Simulated impact force and brain acceleration during the impact. A: Time histories of the impact forces. Dotted and solid line represent the impact of the object-helmet (F_a) and head-helmet suspension (F_b), respectively. B: The maximal acceleration in the brain was observed around 3.0 ms.

proximately 60%, which represents one of the aspects of the helmet performance. The calculated distributions of the acceleration in the brain (Fig. 2B) show that the maximal values were at the occipital region. It is interesting to see that the maximal acceleration was observed around 3 ms, a delay about 0.3 ms from the peak of F_b. The delay of the acceleration peak is likely caused by the viscous effects of the soft tissue materials.

Typical falling objects in construction site are small and have a mass around 2 kg, such as hand tools, bricks, bolts, etc. The mass and dimension of the falling object simulated in our study is representative for real situations. In the simulations, we selected an impact velocity of 8 m/s for the object, which is approximately correspondent to an object falling height of 5 m, assuming a worker has a height of 1.8 m. This height is typical at construction sites of residential buildings in the United States. The purpose of this study is to develop a model; once the model is validated, it can be applied to analyze or to numerically reconstruct accidents at construction sites.

4 FUNDING, CONFLICT OF INTEREST, AND DISCLAIMERS

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The findings and conclusions in this report are those of the authors and do not necessarily represent the views of the National Institute for Occupational Safety and Health. Mention of company names or products does not imply endorsement by the National Institute for Occupational Safety and Health.

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PREDICTING THE EFFECTS OF STRAIGHT AND TAPERED STENTS ON BLOOD FLOW PROPERTIES IN THE AORTA

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SUMMARY

Coarctation of the Aorta is a congenital heart disease with a severe prognosis. Several treatments are available, but all treatments have side effects. Furthermore, since some treatments have only been commonly used in the past two decades, clinical data that could be used to compare the long-term outcomes of the treatments are unavailable.

We simulate blood flow through one-dimensional models of treated aortas to assess the impact of treatments on blood flow properties. In this study, stent treatments are modelled and we find that straight stents, as opposed to stents that taper with the aorta, exhibit preferable blood flow properties.

Key words: *coarctation, aorta, stent, model*

1 INTRODUCTION

A serious congenital heart disease, Coarctation of the Aorta (CoA) is characterised by a narrowing of the aorta, decreased blood flow downstream of the narrowing, and increased blood flow upstream [1]. Treatments that increase the lifespan of patients are available, but treated patients experience a decreased lifespan and an increased incidence of other diseases, such as hypertension and aneurysms [2]. Stent placements have emerged as a viable alternative to surgical techniques [3]; however, since they have been commonly used only since the 1990s, long-term clinical data are unavailable. A stent treatment involves inserting a compressed metal mesh inside the coarctation on a catheter, then expanding the metal mesh until it dilates the coarctation and is embedded in the artery wall.

Computational fluid dynamics has been used extensively to simulate blood flow in various vessels, in both one-dimensional (1D) and three-dimensional (3D) models. Compared to *in vivo* measurements, *in vitro* experiments and 3D models, 1D models have successfully reproduced many features of the fluid dynamics of blood flow in large human arteries like the aorta, including pressure, flow and area waveforms [4].

In this paper we use 1D models of treated aortas. We represent the stent treatments in the aorta by increasing the artery wall stiffness in the region that the stent covers, to determine if the increased stiffness from stents can explain complications, such as hypertension, seen in treated patients. We also model a stent that tapers with the natural tapering of the aorta, as well as a straight stent, to see the effect of the stent shape.

2 METHODOLOGY

The aorta is a thick-walled, elastic vessel that carries blood from the heart. The contraction of the heart produces blood flow velocity and pressure waves that interact with the artery walls [5]. When the heart contracts, there is a sudden increase in blood volume in the aorta. The aorta exhibits compliance; it distends when blood is pumped into it, then contracts, producing a subsequent pulse wave.

The assumptions made in this 1D formulation are: the aorta is compliant; the blood flow is laminar [6], as blood in large arteries has a low Reynolds number [7]; the pulse wave has forward and