11th World Congress on Computational Mechanics (WCCM XI)
5th European Conference on Computational Mechanics (ECCM V)
6th European Conference on Computational Fluid Dynamics (ECFD VI)
E. Oñate, J. Oliver and A. Huerta (Eds)

MULTISCALE MODELING OF BLAST INDUCED TRAUMATIC BRAIN INJURY: FROM WHOLE BODY RESPONSES TO BRAIN MICRODAMAGE

Raj K. Gupta[1*] and Andrzej J. Przekwas[2]

[1] DoD Blast Injury Research Program Coordinating Office, HQ, US Army Medical Research and Materiel Command, 504 Scott Street, Fort Detrick, MD 21702, USA.
raj.k.gupta.civ@mail.mil, mrmc.amedd.army.mil
[2] CFD Research Corporation, 701 McMillian Way NW, Huntsville AL 35806, USA,
ajp@cfdrc.com, www.cfdrc.com

Key Words: *Blast Wave, Traumatic Brain Injury, Mathematical Model, Biomechanics, Multiscale Model*

Abstract. The blast induced Traumatic Brain Injury (bTBI) has become a signature wound of therecent military operations. In spite of immense clinical and preclinical research on TBI, current understanding of injury mechanisms is limited and little is known about the short and long-term outcomes. Unlike the impact-related brain injury, the mechanisms involved in blast induced mild TBI (mTBI) have not been clearly understood. Mathematical models of human body, head and brain responses to a blast wave may provide capabilities to study brain injury mechanisms, perhaps accelerating the development of neuroprotective strategies and aiding in the development of improved personal protective equipment. The paper presents a novel multiscale, multiphysics simulation framework for modeling blast induced brain injury. We identify modeling components needed for detailed analysis of blast wave threat characterization, human body loading, body biodynamic response and body/brain biomechanics leading to potential primary injury. The paper also discusses the need for coupled modeling of primary injury biomechanics, secondary injury mechanobiology and model based assessment of injury severity scores.

1 INTRODUCTION

Blasts injuries from improvised explosive devices (IEDs) are the most common cause of combat casualties in recent military operations and acts of terror on civilian population[1,2,3]. The blast-induced traumatic brain injury (bTBI) has become a signature wound of recent military activities and terrorist acts. A recent RAND report estimates that almost 20% of the deployed force potentially suffers from TBI[4]. In contrast to previous conflicts, in which gunshot wounds were the primary mechanism of trauma and the nature of the injury was focal, the blast injury often results in distributed injury to various organs including extremities, lungs, ears, and most importantly, brain. Although most of mTBI cases are expected to recover, persistent symptoms after injury, such as chronic dizziness, fatigue, headaches and delayed recall of memory are common[5,6]. The mechanisms of bTBI are not fully understood, and as a result it is difficult to develop personal protective equipment, helmets in particular, that could protect against bTBI.

In spite of immense research on impact-related brain injury due to vehicle crash and sport injuries in civilian population, current understanding of injury mechanisms is limited and little is known about the short- and long-term outcomes of mTBI. Unlike the impact-related brain injury, the mechanisms involved in blast induced mTBI have not been clearly understood. Over the last few decades the Department of Defense (DoD) has performed substantial research on blast trauma to the body, primarily to address injuries seen in previous conflicts and to improve personal protective equipment (PPE)[7]. The resulting improvements in the PPE and trauma care have mitigated or reduced potential blast and ballistic injury to the thorax but vulnerability to face, ear, brain, groin and extremity injury still remain[8]. Protection against blast wave TBI is particularly challenging because, in spite of the protective helmet, a significant part of the soldier's head is still exposed to the blast. Until recently, it was not clear how a blast wave penetrates the cranium and causes brain injury and, if and how military helmets protect against it[9-11]. Better understanding of the blast wave injury mechanisms may be possible with a complementary experimental and computational modeling approach. Validated biomechanics and physiology based mathematical modeling tools of blast head injury may not only reduce the need for trial-and-error tests involving laboratory animals, but also provide a capability to study brain injury mechanisms, perhaps accelerating the development of prevention and mitigation strategies [12-14].

Mathematical models of brain injury biomechanics have been developed for decades, primarily to study accidental impacts and vehicle crashes [15-18]. Models of explosive blast TBI are not well established yet because the injury mechanisms are not well understood and the computational methods needed to simulate these fast and multiphysics events are inadequate. The goal of this paper is to present an example computational framework for multiscale modeling of blast wave TBI integrating various modeling aspects including human body anatomical geometry, blast wave physics, human body biodynamics and body/brain injury biomechanics.

2 MULTISACLE MODELING FRAMEWOK OF BLAST TBI

A comprehensive computational framework for modeling blast injury may involve several disciplines including: human body anatomic geometry, blast wave gas dynamics and body loading, body dynamics and body/head/brain biomechanics. The complexity of such a modeling framework is magnified by a wide spectrum of length scales and time scales[14].

2.1 Anatomy/geometry/mesh

Simulations of blast wave interaction with a human body requires both, a 3D geometry of a human body "immersed" in a computational domain used for computational fluid dynamics (CFD) blast wave physics as well as 3D anatomic geometry of the body internal organs/tissues for finite element (FEM) biomechanical simulations. Until recently brain injury models focused on blunt injury for which the skull and brain geometry were sufficient[15,18, 19]. Since the blast loads have a spatial and temporal distribution over the entire head and neck, the anatomical model should include the head's skin, facial structures including ocular and nasal cavities, cranium, and neck geometry. According to recently reported experimental animal tests, the blast wave loads on the entire body, not just the head, should be included to account for various injury mechanics such as head movement of thoracic vascular waves[20,14].

In the presented computational framework a high resolution of the whole human body anatomic geometry was established based on medical imaging data from the visible human[21] and anatomic databases[22-25]. The anatomic model includes the skin, entire skeleton, all the vital organs/tissues and the vascular system, Figure 1. Two computational meshes were generated for modeling blast injury: a) external body surface adaptive Cartesian mesh for blast CFD simulations, Figure 1, and b) hexahedral mesh of the whole body anatomy for FEM biomechanics simulations. Typical mesh sizes used for such simulations involve approx. 15 million cells in the blast wave CFD domain and 4-5 million elements inside the human body for FEM biomechanics.

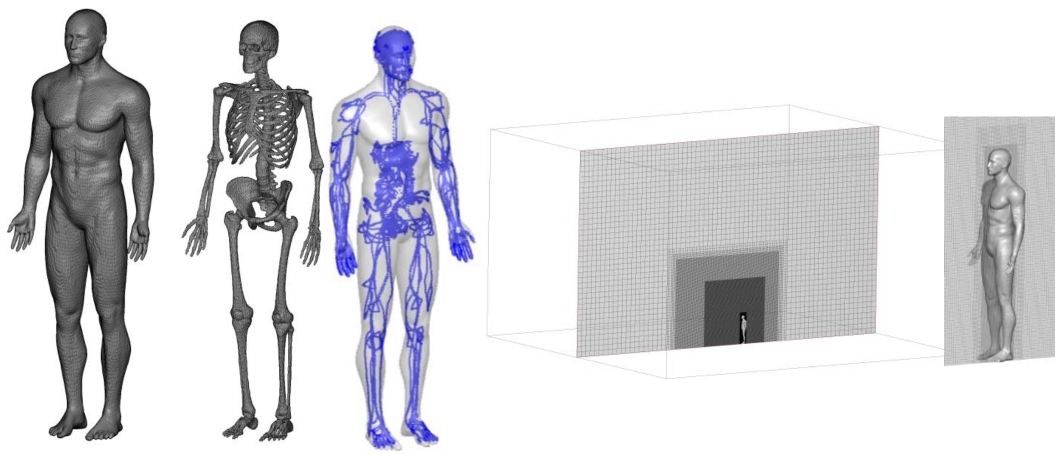

Figure 1. Anatomical geometry model of a human body and a computational mesh for CFD blast wave dynamics and body loading simulations.

2.2 CFD model of blast wave dynamics and body loading

Computational modeling of an IED explosion involves two main events: initial charge detonation and subsequent propagation of a blast wave and its interaction with objects. Buried IED will also generate high speed flying debris and ejecta. Because of complexity of the detonation physics, typical CFD simulations start with assumed initial blast wave shape and properties (pressure, temperature, volume) and propagate the wave towards the human body[26-29]. The initial "post detonation" properties can be calculated analytically or by using semi-empirical models such as ConWep. Figure 2 presents an example simulation results of a blast wave initiated by a detonation of 5lb C4 located 1.27m above the ground at a distance of 2.33m from a human body. The non-reflecting outlet boundary condition is applied at the boundaries of the computational domain. To reduce the computational burden a sequential strategy was developed using a very fine mesh 1D spherical model for the initial explosion stage leading up to the wave contact with the ground, followed by 3D simulations of the wave interaction with the ground and the human body. The accuracy of the 1D initial explosion model has been verified with the semi-analytical curve fit results[23]. Reported simulations have shown that such a sequential modeling approach is well justified as the inertial body movement starts well after the blast wave traverses the body[29,24]. The CFD simulations provide time dependent pressure and shear force loads on the entire human body surface. These forces are then used as loading conditions for subsequent human body dynamics and biomechanics simulations. Figure 2 also shows pressure profiles on the human body at selected time instances during the blast wave exposure. As expected, the highest loads are on the body surface directly facing the blast front. Observed reflected pressures on the body surface were much higher than the free shock wave overpressure and the highest pressures are located on the concave body cavities such as eye sockets, ears, lower neck, groin and between the thighs.

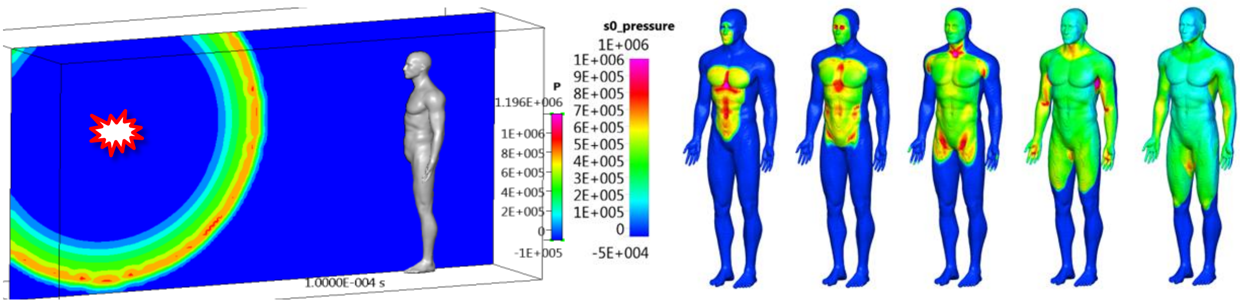

Figure 2. CFD computational results of a blast wave explosion above the ground and blast wave induced pressure loads on a human body at several time instances.

2.3 Human Body Biodynamic Response to Blast Loads

The same anatomical body can also be used to simulate body biodynamics (movement, translocation impact on the ground). An articulated body model, shown in Figure 3, includes several rigid body segments “connected” with joint interfaces. The time dependent blast loads on the skin are transformed as forces and moments on each body segment and used as inputs for body biodynamic simulations. To study human body dynamics (translocation in air and impact on the ground) we used three loads on the body: (1) time dependent blast wave pressures distributed on the entire body surface, (2) gravity force acting on the whole body, and (3) reaction forces between body and the ground due to impact and friction. The blast forces are undoubtedly the biggest but last only for the first few tens of milliseconds. After that the gravity forces dominates until body contacts with the ground. The ground contact forces can be considerable since they are localized on small areas on the injury sensitive body including head or pelvis. Details of the model and model validation have been previously reported[22-25].

The articulated human body simulation results are shown in Figure 3 for the body movement at several time instants. After the blast wind has passed at around 6.5 milliseconds, the human body does not move much (Figure 3, second silhouette) and the maximum displacement in the whole body is less than three centimeters, which justifies the one-way coupling strategy for the simulation of blast-human body interaction. The pelvis hits the ground in approx. 0.95sec and the head contacts the ground in 1.2sec. Small asymmetries of the body anatomy and the blast wave cause the marked asymmetries in body moment and impacts. Using the predicted head acceleration history one can assess the probability of head injury according to the acceleration-based head injury criterion HIC.

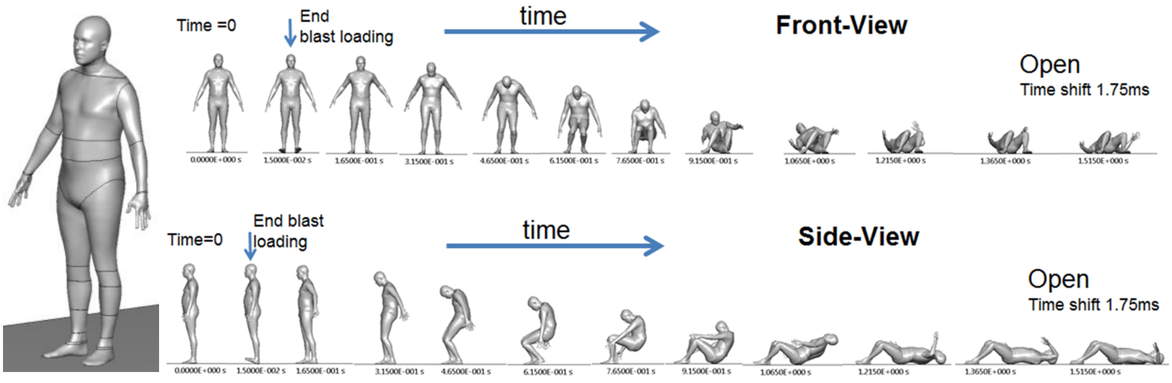

Figure 3. Articulated human body model and several time instances from human body biodynamic response to a front exposure blast load.

2.4 Human Body and Brain Biomechanics in Response to Blast Loads

The blast loads are also used to simulate propagation of pressure and shear waves inside the human body. The anatomic geometry of the body, Figure 1, is represented with hexahedral mesh conforming to the skin and selected organs such as brain, spine, lung, liver with mechanical properties calculated based on underlying tissues. For example, the lungs are modeled as separate organs because the sound speed in the lung is an order of magnitude slower compared to other tissues. The brain and spinal cord are modeled as an isotropic viscoelastic material, without considering the difference between white mater and grey mater. The cerebrospinal fluid (CSF) layer between the skull and the brain is not explicitly modeled but is considered to be part of the brain. The typical element size is approximately 2.5mm and the total number of elements in the whole body is over 4.2 million. An explicit FEM solver module in in CoBi framework is used to simulate internal body biomechanics.

The blast induced body/brain biomechanics model has been validated in experimental data for a rat in a shock tube[25] and on a surrogate head in a shock tube[23]. Figure 4 presents pressure fields within the body at three time instances, pressure loads on the skeleton and on the brain and the spinal cord. It can be noticed that stiffer material like the skeleton has the higher pressure, while the pressure in the soft material like the lung is lower. CoBi FEM biomechanics simulations generate the pressure time history in all organs including the vasculature. That pressure data is then used as the loading condition for the vascular injury model[30].

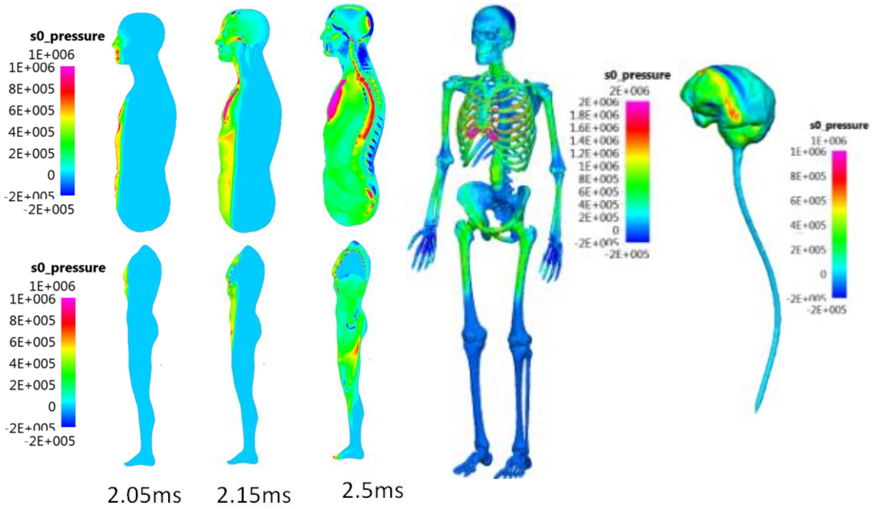

Figure 4. FEM simulation results of a pressure filed in the body at three selected time instanced during pressure wave propagation, blast loads on the skeleton and on the brain.

2.5 Brain Injury Mechanisms

There are several potential pathways for the blast wave energy to enter the brain, including: (1) the skull deformation creating a stress wave within the brain, (2) translation/rotation of the head causing compression/shear waves within the brain as well as brain rotation within the skull, (3) the pressure wave directly entering the brain via various foramina (orbital, ethmoidal, vestibulo-cochlear, foramen magnum, and vascular foramina), and (4) an elastic wave propagating along blood vessels from a compressed thorax[14]. A validated computational model of brain tissue biomechanics will not only provide detailed macroscopic time-space resolved stress/strain information but may also generate loading conditions for the brain microstructures including neuronal axons, dendrites, synapses and the blood brain barrier. The need and benefits from such multiscale models have been recently identified[14,31] and computationally demonstrated[32].

FEM biomechanics models may also be used to simulate brain tissue/cell damage at the micro-scale. Mathematical models of mechanical damage to neuroaxonal structures may be able to describe damage to cell membranes, cytoskeleton, ion channels, synaptic clefts, dendrites and axons. These in turn, could provide inputs for the secondary injury and repair models, simulating electrophysiology and ion homeostasis, alterations in metabolism, neuroexcitation, cytotoxic edema, oxidative stress, apoptosis and other injury and repair mechanisms. In the last few years, the first FEM biomechanics simulations of very simplified axonal structures have been reported[32-34]. Future advancements in micro-scale FEM can incorporate boundary conditions from macro-scale simulations[32-37]. *In vivo* micro-imaging may also provide functional response data (electrophysiological, metabolic, and biochemical) needed for the development and validation of mathematical models of secondary brain injury and repair mechanisms. We envision that the next generation of *in vivo* and *in vitro* micro-biomechanics models will be able to elucidate neuroaxonal injury mechanisms and will help establish brain region and insult specific injury criteria.

3 CONCLUSIONS

Current state of the art models of blast waves and head/brain biomechanics provide an excellent foundation for the development of a primary brain injury model at the macro- and micro-scale. The most challenging step is to link the models of the primary blast event with the resulting brain tissue damage including the secondary mechano-biology of injury to the neuro-functional outcome. Such a multi-scale multi-physics model requiring a concerted collaborative effort between biophysicists, neurobiologists, mathematicians and experimentalists could play a major role in advancing our understanding of brain injury mechanisms, and help in neurodiagnostics, treatment and protection.

ACKNOWLEDGEMENTS

We would like to thank colleagues at CFDRC for stimulating discussions and contributions to CoBi TBI model development and to Dr. Tan for providing simulation results.

DISCLAIMER

The views expressed in this paper are those of the authors and may not necessarily be endorsed by the U.S. Army or U.S. Department of Defense.

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