In Situ Intracranial Strain Measurements within an Elastomeric Brain Surrogate

by

Jennifer Rovt

A Thesis submitted to
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the requirements for the degree of
Master of Applied Science

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Mechanical and Aerospace Engineering

Department of Mechanical and Aerospace Engineering
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The undersigned recommend to
the Faculty of Graduate Studies and Research
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Brain Surrogate

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Carleton University
2022
Abstract

Incidence of concussion remains high despite the widespread use of helmets. While the primary cause of mild traumatic brain injury (mTBI) is thought to be intracranial strain, current helmet evaluation techniques resolve head kinematics as the primary evaluation metrics. These techniques have been highly effective in reducing focal brain injuries. However, their effectiveness in reducing the incidence or severity of concussion has been less clear. There remains a need to advance tools and methodologies capable of making a more direct link between helmet certification protocols and the causes of concussive injury. This study presents displacement and strain within a deformable head surrogate, the BIPED headform, subjected to an extensive series of impacts. Impacts were captured under high speed X-ray at 5,000 fps, and strain fields were computed using digital image correlation. Results from this study were compared to cadaveric brain tissue displacements measured under similar impact experiments.
Acknowledgments

As I reflect on my time at Carleton, I’d like to take a moment to acknowledge the incredible group of people who have made this chapter in my life possible.

Firstly, this work would not have been possible without the knowledge and support of my supervisor, Dr. Oren Petel. You have pushed me to grow in ways unimaginable, and with your guidance and mentorship, I have found a space where I truly feel excited about the work that I have, and will continue to, put forward.

This work was funded by the Canadian Institute of Health Research and the Natural Sciences and Engineering Research Council through the Collaborative Health Research Program (CPG-151967). The equipment used in this study was funded by the Canada Foundation for Innovation and the Ontario Research Fund (Project 32933).

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To my past and present lab-mates, your advice, conversations, idea sounding boards, and presentation feedback were all greatly appreciated. In particular, I must thank Sheng Xu and Scott Dutrisac, with whom I have had the immense privilege of working closely over the past two years. You have both taught me more than you know, and this work would not have been possible without you. I must also single out Hannah Thomson and Ashley Mazurkiewicz, who have been there for endless conversations about research, and the subsequent distractions from it. I have been very fortunate to have you both as colleagues and good friends.

I would also like to thank the support and technical staff in the Carleton University Department of Mechanical and Aerospace Engineering, whose advice and assistance in manufacturing and experimental work, and guidance through administrative matters was invaluable.

Lastly, this chapter in my life would not have begun if not for the support of those closest to me. To my parents, Val and Alex; and my sister, Renata: you have always encouraged me to take on new challenges in the pursuit of my passions, and for that I am eternally grateful. To my partner, Simon, thank you for being by my side throughout this entire journey. You make every obstacle easier to overcome, every moment of uncertainty steadier, and every success, sweeter.
Preface

The work presented in this thesis was performed at the Carleton University Impact Research Laboratory with collaboration from Defense Research and Development Canada Valcartier, and the Neurotrauma Impact Science Laboratory at the University of Ottawa.

Ethics Approval

Cadaveric specimens for the PMHS portion of this work were obtained through the University of Ottawa Body Bequeathal Program. This work was approved by the Carleton University Research Ethics Board (REB #107190).

Contributions

The high-speed X-ray imaging system used in this work was developed and implemented by Professor Oren Petel with assistance from Stephane Magnan, Scott Dutrisac, Karen Taylor, Ashley Mazurkiewicz, Sheng Xu, and MacKenzie Brannen.

The BIPED Headform was designed and developed by Defense Research and Development Canada Valcartier, with modifications by Sheng Xu. The marker design was conceived by Scott Dutrisac and manufactured and implemented into the BIPED by Sheng Xu. The head surrogate impact experiments were developed and carried out by Jennifer Rovt with assistance from Sheng Xu and Scott Dutrisac. Image processing and video stabilization methods were developed in MATLAB by Scott Dutrisac and performed by Jennifer Rovt. Digital image correlation was performed by Jennifer Rovt with assistance from Sheng Xu, using an open-source
MATLAB program, NCorr. Analysis and CORA analysis was performed by Jennifer Rovt.

The tools and methods used for image processing, video stabilization, particle tracking, and displacement measurement in the PMHS portion of this study were developed in MATLAB by Scott Dutrisac and performed by him and Jennifer Rovt. Details of the methodology can be found in A Study of the Impact Response of Discrete Regions of the Human Cadaver Brain (Dutrisac 2020). The PMHS impact protocol was performed by Scott Dutrisac and Jennifer Rovt with laboratory assistance from Professor Petel, Dr. Andrew Post, Mary-Jo Weir Weiss, MacKenzie Brannen, Gia Kang, and Klara Doelle. Specimens were prepared at the University of Ottawa by Prosector Shannon Goodwin. MRI imaging was performed at The Royal Ottawa Hospital by Dr. Greg Cron, Dr. Gerd Melkus, and Dr. Reggie Taylor.

Much of this work has been submitted for consideration to Scientific Reports and is currently under review under the following information:

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# Nomenclature

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<thead>
<tr>
<th>Symbol</th>
<th>Description</th>
<th>Units</th>
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<tbody>
<tr>
<td>$a$</td>
<td>Linear acceleration</td>
<td>$[m/s^2]$</td>
</tr>
<tr>
<td>$E$</td>
<td>Strain</td>
<td>$[- -]$</td>
</tr>
<tr>
<td>$g$</td>
<td>Linear acceleration</td>
<td>$[9.81 m/s^2]$</td>
</tr>
<tr>
<td>$I$</td>
<td>Moment of inertia</td>
<td>$[Nms^2]$</td>
</tr>
<tr>
<td>$t$</td>
<td>Time</td>
<td>$[s]$</td>
</tr>
<tr>
<td>$u$</td>
<td>Local $x$-displacement</td>
<td>$pixel$</td>
</tr>
<tr>
<td>$v$</td>
<td>Local $y$-displacement</td>
<td>$pixel$</td>
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## Greek Characters

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<tr>
<th>Symbol</th>
<th>Description</th>
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<tbody>
<tr>
<td>$\alpha$</td>
<td>Angular acceleration</td>
<td>$[rad/s^2]$</td>
</tr>
<tr>
<td>$\omega$</td>
<td>Angular rate</td>
<td>$[rad/s]$</td>
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### List of Abbreviations

<table>
<thead>
<tr>
<th>Abbreviation</th>
<th>Description</th>
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<tbody>
<tr>
<td>ABS</td>
<td>Acrylonitrile butadiene styrene</td>
</tr>
<tr>
<td>ATD</td>
<td>Anthropometric testing device</td>
</tr>
<tr>
<td>BaSO4</td>
<td>Barium sulfate</td>
</tr>
<tr>
<td>BIPED</td>
<td>Brain Injury Protection Evaluation Device</td>
</tr>
<tr>
<td>BrIC</td>
<td>Brain Injury Criteria</td>
</tr>
<tr>
<td>BSI</td>
<td>British Standards Institution</td>
</tr>
<tr>
<td>CDSM</td>
<td>Cumulative strain damage measurements</td>
</tr>
<tr>
<td>CG</td>
<td>Centre of gravity</td>
</tr>
<tr>
<td>CSF</td>
<td>Cerebrospinal fluid</td>
</tr>
<tr>
<td>CT</td>
<td>Computed tomography</td>
</tr>
<tr>
<td>DAMAGE</td>
<td>Diffuse Axonal Multi-Axis General Evaluation</td>
</tr>
<tr>
<td>DIC</td>
<td>Digital image correlation</td>
</tr>
<tr>
<td>DRDC</td>
<td>Defense Research and Development Canada</td>
</tr>
<tr>
<td>DSNM</td>
<td>Dartmouth Scaled and Normalized Model</td>
</tr>
<tr>
<td>FE</td>
<td>Finite element</td>
</tr>
<tr>
<td>FPS</td>
<td>Frames per second</td>
</tr>
<tr>
<td>FS</td>
<td>Full scale</td>
</tr>
</tbody>
</table>
GHBMG Global Human Body Models Consortium
GSI Gadd Severity Index
HARM Head Acceleration Response Metric
HIC Head Injury Criterion
HIP Head Impact Power
HPS Helmet Performance Score
HSXR High-speed X-ray
ISO International Organization for Standardization
KTH KTH Royal Institute of Technology
MPS Maximum principal strain
MRI Magnetic resonance imaging
mTBI Mild traumatic brain injury
NDT Neutral density target
NFL National Football League
NHTSA National Highway Traffic Safety Association
NOCSE National Operating Committee on Standards for Athletic Equipment
PDMS Polydimethylsiloxane
PLA Polylactic acid
PMHS Post-mortem human surrogate
<table>
<thead>
<tr>
<th>Abbreviation</th>
<th>Description</th>
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<tbody>
<tr>
<td>PMMA</td>
<td>Poly methyl methacrylate</td>
</tr>
<tr>
<td>PRHIC</td>
<td>Power Rotational Head Injury Criterion</td>
</tr>
<tr>
<td>RIC</td>
<td>Rotational Injury Criterion</td>
</tr>
<tr>
<td>ROI</td>
<td>Region of interest</td>
</tr>
<tr>
<td>SIMon</td>
<td>Simulated Injury Monitor</td>
</tr>
<tr>
<td>TBI</td>
<td>Traumatic brain injury</td>
</tr>
<tr>
<td>THUMS</td>
<td>Total Human Model for Safety</td>
</tr>
<tr>
<td>VN</td>
<td>Vinyl nitrile</td>
</tr>
<tr>
<td>WSUHIM</td>
<td>Wayne State University Head Injury Model</td>
</tr>
<tr>
<td>XDIC</td>
<td>X-ray digital image correlation</td>
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Chapter 1

Introduction

1.1 Traumatic Brain Injury

Traumatic brain injury (TBI) is defined by the National Institute of Neurological Disorders and Stroke, and the National Institute on Disability and Rehabilitation Research as “an alteration in brain function, or other evidence of brain pathology, caused by an external force” [3, 4]. TBI refers to a broad spectrum of injury, and is often classified as one of two principal mechanisms: focal brain injury and diffuse brain injury [5, 6].

Focal brain injury occurs as a result of collision forces acting on the skull, and has been linked to linear accelerations of the head [5, 7]. Tissue underneath the skull is compressed either at the site of impact (coup), or opposite the site of impact (contre-coup) resulting in contusions, lacerations, subdural and epidural haematomas, and intracranial haemorrhage [5, 6]. Diffuse brain injury occurs when tissue segments with varying densities cause differing inertial properties in the brain, resulting in the strain of brain tissue [8, 9]. This may lead to diffuse axonal injury, diffuse vascular injury, hypoxic-ischemic injury, and brain swelling [5, 6].

While diffuse injuries generally tend to be more difficult to diagnose through traditional imaging such as computed tomography (CT) or magnetic resonance...
imaging (MRI); in its milder state, mild traumatic brain injury (mTBI) or concussion may cause symptoms in the absence of positive neuroimaging findings altogether [5, 10]. As there is a lack of reliable biomarkers and no known mechanical threshold for clinical concussions, diagnosis remains a challenge and is often based on self-reported neurological symptoms [4, 10, 11], with as many as 50% of concussions going unreported [11].

1.2 Motivation

Despite the challenges associated with mTBI diagnosis, an annual average of 147,815 Ontarians (1,153 per 100,000 residents) were diagnosed with concussion between 2008 and 2016 [12]. In the United States it is estimated that 3.8 million concussions occur during competitive sports and recreational activities annually, with 30% of all concussions in individuals aged 5-19 years being sport-related [11]. Given the high incidence of mTBI, there is a clear motivation to implement preventative measures and reduce the occurrence of injury. In the case of sport-related mTBI, these preventative measures are often in the form of protective equipment, such as helmets.

The use of helmets, to date, has been highly effective in preventing focal injuries, such as skull fracture, laceration, and hemorrhage [11, 13]; however, their success in reducing the incidence or severity of concussion remains uncertain [14, 15]. It is hypothesized that this is due to the primary metrics used in helmet certification, resulting in an emphasis on the reduction of resultant linear head acceleration and related injury severity scales.

Current helmet certification methodologies involve subjecting helmet-equipped anthropometric testing devices (ATD; often referred to as headforms)
to a series of tests, including guided drops, linear and projectile impacts, and fit-related tests \cite{16-18}. While there are numerous headforms in use across the various standards with ranging levels of biofidelity \cite{19-21}, their measurement and evaluation approaches are similar. The head surrogates measure the loading and kinematic responses to these impacts using an array of sensors. Helmet performance is evaluated using peak kinematic values and severity indices \cite{18}, which link kinematic responses to injury outcomes \cite{22-24}. While these approaches provide a broadly accessible testing methodology for injury prediction, the measured outcomes do not provide a direct link between the test conditions and brain tissue damage.

As earlier studies have identified tissue damage in animal models at shear strain rates greater than 10 s\(^{-1}\) \cite{25}, strain is broadly hypothesized to be the leading mTBI injury mechanism \cite{9}. Given that repetitive head trauma and mTBI have been linked to neurodegenerative diseases with devastating health outcomes \cite{26,27}, there is a need to advance testing methodologies capable of providing a direct link between mTBI mechanisms and helmet performance evaluation.

### 1.3 Objectives

The work described in this thesis aims to present and evaluate a preliminary iteration of a novel ATD capable of directly measuring intracranial strain during impact loading conditions. This tool will assist in bridging the gap between the causes of concussive injury and helmet performance metrics. The headform used in the present study was originally designed by Defense Research and Development Canada Valcartier to investigate blast-induced traumatic brain injury. It has been calibrated against cadaveric datasets under those loading conditions \cite{28,29}. The process of testing early iterations of the headform and performing verification
and validation testing has played a role in the broader development of this tool and is the primary focus of the results presented in this work.

The specific goals of this thesis were:

- To present a novel head surrogate and helmet evaluation methodology based on measuring intracranial strain using high-speed X-ray digital image correlation (XDIC); and
- To present preliminary verification and validation findings to further guide the design and development of future iterations of the BIPED headform.

1.4 Thesis Overview

This work is organized into several chapters, described below, comprising background information, descriptions of experimental methodologies, and a presentation and preliminary validation of results.


Chapter 2: Helmet Testing and Head Injury Prediction — A review of current helmet testing standards and helmet evaluation methodologies, with a focus on injury prediction modelling, deformable headforms, and intracranial strain estimation tools.

Chapter 3: Experimental Methodology — Description of the methods used for experimental data collection and analysis of intracranial strain measured within the BIPED headform.
Chapter 4: Results — Presentation of tissue-surrogate displacements and intracranial strains, as well as error investigation and measurement repeatability results.

Chapter 5: Discussion — Discussion of the advantages of this technique and insights gathered from the intracranial strain measurements, followed by a discussion on headform validation and calibration.

Chapter 6: Concluding Remarks — Summary of key findings and recommendations for future work on the BIPED headform.
2.1 Headforms and Helmet Testing

A number of standard and certification bodies release specifications for the evaluation and certification of headgear. While many standards have been enacted to encompass the numerous helmet use-cases, many of the standards borrow similar requirements and helmet testing protocols.

The earliest helmet standards were published by the British Standards Institution (BSI), beginning with BS 1869-1952 Crash Helmets for Racing Motorcyclists, in 1952 [30,31]. Following this standard, the BSI published helmet standards for racing car drivers, motor cyclists, and light and heavy duty industrial safety helmets [31]. These earliest standards measured the dynamic force resulting from a 10 lb wood block dropped onto a helmet-equipped wooden headform from a height of 9 ft, with a failure criterion of 5000 lb-f (almost 500 g) [30,31]. The wooden headform had minimal specifications, mostly relating to the manufacturing of the headform rather than any emphasis on biofidelity [30].

The first American sports helmet standards were developed by Snively for racing crash helmets and published through the Snell Memorial Foundation in 1959 [30]. This standard was the first to measure acceleration directly, using a
12 lb freely moving K1-A magnesium alloy headform, with a failure criterion of 400 g [30]. While other parameters for this headform were not specified, the American Standards Association (later American National Standards Institute) provided a similar specification with ASA Z90.1-1966, Protective Headgear for Vehicular Users, which added additional metrics for biomechanical tolerance levels using the same style of headform (Figure 1b) [30]. This standard maintained the 400 g failure criterion, and added that acceleration in excess of 200 g must not persist for longer than 2 ms, and accelerations in excess of 150 g must not persist for longer than 4 ms [30].

In 1983, the International Organization for Standardization (ISO) released draft standard ISO/DIS 6220-1983 Headforms for Use In the Testing of Protective Helmets [31]. Though it was never adopted as an international standard, all major voluntary standards in the United States for the evaluation of cycling helmets refer to this draft standard when establishing their K1-A magnesium alloy headform dimensions [32]. In addition to cycling helmets, magnesium alloy headforms remain in use for a number of modern helmet testing standards, including the testing of ice hockey helmets, mountaineers helmets, and ski helmets, despite their minimal biofidelity [33–36].

Football helmets took a different approach with the founding of the National Operating Committee for Athletic Equipment (NOCSAE) in 1969. The most significant departure from existing standards was the development of a test headform that based its anthropometry on cadaveric measurements [38]. The headform (Figure 1b), which had a silicone skin simulant, a self-skinning urethane foam skull, and a silicon gel cranial cavity, was designed with a concerted effort towards biofidelic geometry and inertial response to impact [30, 38]. Unfortunately, the NOCSAE headform has been prone to fracture under severe impact conditions, which
Figure 1: Headforms used in helmet testing. (a) K1-A magnesium alloy headform, reproduced from Cadex Inc. [37] (b) NOCSAE headform [38], reproduced from Cobb [19], and (c) Hybrid III headform [39], reproduced from Cobb [19].

offers similarity to the human skull, but is problematic in the context of a robust piece of testing equipment [30,40].

NOCSAE has since developed certification standards for a number of sporting helmets, including baseball, lacrosse, ice hockey, and polo [41], which involve extensive testing protocols. In addition to drop tests measuring linear acceleration conducted with the NOCSAE headform [17], modern NOCSAE helmet standards also evaluate the response to projectile impacts using the NOCSAE headform [42], and the response to linear impacts using a Hybrid III headform [18].

The Hybrid headform was developed by Hubbard and McLeod through a General Motors Anthropomorphic Dummy Development Program, and first used in 1972 [20,39,43]. The current model of the headform, the Hybrid III, is used extensively on National Highway Traffic Safety Administration (NHTSA) crash test dummies [44], as well as all linear impact tests conducted by NOCSAE in their helmet evaluations [18], in addition to commonly being used in head impact research [45-47]. The Hybrid III headform is constructed from a cast aluminum skull
which houses instrumentation (such as an accelerometer), and a vinyl skin [20][39]. The headform size, shape, and weight are representative of a 50th percentile adult male, and it features an anatomically correct center of gravity (CG) location and head-to-neck interface [43]. The headform (Figure 1a) was designed to simulate the forehead impact response for flat, rigid surface impacts based on cadaver response data collected by Hodgson and Thomas [48]. An analysis by Mertz on the biofidelity of the Hybrid III head found the response to be adequately representative for this head impact configuration, with excellent response repeatability and mid-sagittal plane symmetry, as well as excellent durability [43]. However, since the Hybrid III was designed for frontal impacts in particular, the headform’s mass moment of inertia has only been specified along the lateral axis and may not be suitable for impact tests where the response along all axes are necessary [40].

2.2 Models for Head Injury Prediction

To better contextualize the acceleration data obtained during head impacts and helmet testing, a number of kinematic and computational models have been developed to link kinematic data to risk of head injury and clinical outcomes. These models of head injury prediction have become very intertwined with helmet testing standards, with many of the standards using modeling to establish failure criteria and inform future helmet designs.

One of the earliest seminal models, the Gadd Severity Index (GSI), assigns a score based on both the intensity and duration of the loading on the head using an integral of the following form [49]:

\[ G.S.I. = \int_{T_1}^{T_2} a^n dt \]  

(1)
where \( a \) is an acceleration, force, or pressure response function, \( n \) is a weighting factor, and \( T = T_2 - T_1 \) is the exposure time in seconds. This model proved to be very insightful for the reduction of focal brain injuries, which have been directly linked to linear accelerations of the head [5][7]. It remains in use in a number of helmet testing standards [17][18][33], including the NOCSAE helmet standards.

Building on the GSI model, the Head Injury Criterion (HIC) evaluates the likelihood of a linear head acceleration to result in injury using the following expression [50]:

\[
HIC = \max \left[ \frac{1}{t_2 - t_1} \int_{t_1}^{t_2} a(t) \, dt \right]^{2.5} (t_2 - t_1)
\]

(2)

where \( t_2 \) and \( t_1 \) are any two times during the acceleration pulse, measured in milliseconds, and acceleration is measured in g [50]. This model has been adopted as a pass/fail criterion by NHTSA to evaluate the effectiveness of occupant protection systems in motor vehicles subjected to collisions [50]. In its current application in NHTSA, HIC is specifically used over a 15 ms time interval [51].

Though the HIC model was most notably adopted for its use in the assessment of motor vehicle crashworthiness, it served as the basis for a number of modern injury risk assessment models [23][52] which sought to incorporate additional rotational kinematic metrics to encompass our developing understanding of diffuse brain injuries [53-55], often verified using finite element (FE) head models. For example, a study by Kleiven et al. [23], found that a linear combination of peak change in rotational velocity and HIC was highly correlated with maximum principal brain strain calculated by the KTH Royal Institute of Technology (KTH) FE head model. Kimpara and Iwamoto proposed two models, the Rotational Injury Criterion (RIC) and the Power Rotational Head Injury Criterion (PRHIC) [52], for evaluating risk of TBI based on rotational acceleration. RIC takes a similar form to
HIC, substituting the linear acceleration component in HIC for resultant angular acceleration. PRHIC incorporates the use of the rotational component of the Head Injury Power (HIP) model developed by Newman et al. which considers the rate of change of kinetic energy rather than an acceleration pulse \[52,56\]. These models are given by the following equations:

\[
RIC = \max \left[ \frac{1}{t_2 - t_1} \int_{t_1}^{t_2} \alpha(t) dt \right]^{2.5} (t_2 - t_1) \tag{3}
\]

\[
PRHIC = \left[ (t_2 - t_1) \left\{ \frac{1}{t_2 - t_1} \int_{t_1}^{t_2} HIP_{rot} dt \right\}^{2.5} \right]_{\max} \tag{4}
\]

where the HIP term is given by:

\[
HIP_{rot} = \sum I_{ii} \alpha_i \int \alpha_i dt \tag{5}
\]

and \( i \in \{x, y, z\} \), \( I_{ii} \) is the moment of inertia of the head, and \( \alpha_i \) is the rotational acceleration of the head \[52\]. However, when verified against the Total Human Model for Safety (THUMS) brain model (Figure 2a), it was found that neither of these injury models correlated with injury predictors given by the FE model \[52\].

**Figure 2:** Finite element head models used in the development of various kinematic injury predictors: (a) THUMS head model \[57\], reproduced from Kimpara and Iwamoto \[52\], (b) SIMon head model \[58\], reproduced from Ji et al. \[59\], and (c) GHBMC head model \[60\], reproduced from Mao et al. \[60\].
Other models relied more heavily on strain calculated by FE models to inform the development of kinematic-based metrics. Takhounts et al. created the Brain Injury Criteria (BrIC) model using the Simulated Injury Monitor (SIMon) and the Global Human Body Models Consortium (GHBMC) FE head models (Figure 2b and 2c) [24]. Maximum principal strain (MPS) and cumulative strain damage measurements (CDSM) were correlated to angular velocity through scaled animal injury data and pendulum tests on rigid ATDs, giving the following relationship:

\[
BrIC = \sqrt{\left(\frac{\omega_x}{\omega_x C}\right)^2 + \left(\frac{\omega_y}{\omega_y C}\right)^2 + \left(\frac{\omega_z}{\omega_z C}\right)^2} 
\]

where \(\omega_x\), \(\omega_y\), and \(\omega_z\) are the maximum rotational velocities about the x-, y-, and z-axes [24]. \(\omega_x C\), \(\omega_y C\), and \(\omega_z C\) are critical angular velocities, dependant on the headform used to measure the kinematic response [24]. The correlation between MPS and CDSM with BrIC is shown in Figure 3. Due to the limited availability of human injury data, this kinematic model was limited to animal injury data for validation [24].

![Figure 3: Plots depicting the correlation between (a) BrIC and MPS, and (b) BrIC and CDSM for all headforms, reproduced from Takhounts et al. [24].](image)
The recently developed Diffuse Axonal Multi-Axis General Evaluation (DAMAGE) model provides a rapid estimation of the 95\textsuperscript{th} percentile peak MPS, as calculated by GHBMC head model \cite{60,61}. The DAMAGE model uses a mechanical analogue - a second-order mass-spring-damper system (Figure 4) - to represent the maximum intracranial strain resulting from directionally dependent head impact angular acceleration profiles \cite{61}. To calculate the DAMAGE parameter, the equations of motion for a damped, 3 degree-of-freedom mechanical system where the masses are subjected to a forced excitation are solved \cite{61}. The solution, $\ddot{\delta}$, is assumed to correlate to brain deformation resulting from directionally dependent rotational motion of the head. $\ddot{\delta}$ is used to calculate DAMAGE using a scale factor, $\beta$, to relate the maximum resultant displacement to MPS in the GHBMC FE model \cite{61}:

$$DAMAGE = \beta \max_t \{ |\ddot{\delta}(t)| \} \quad (8)$$

**Figure 4:** Second-order mass-spring-damper system used in the development of the DAMAGE model. Reproduced from Gabler et al. \cite{61}.
The DAMAGE model has also been combined with HIC to form the Head Acceleration Response Metric (HARM), developed by Bailey et al. [45], to predict injury using both translational and rotational head kinematics [45]. An extensive review of concussive impacts from the National Football League (NFL) informed headform test conditions to develop an in-laboratory test procedure to rank overall helmet performance. The HARM metric informs the severity of a particular impact, whereas the helmet performance score (HPS) provides a weighted sum of the HARM scores for a particular helmet across the 18 test conditions (6 impact locations, at 3 impact speeds each) (Figure 5) [45]. The HARM and HPS metrics are calculated by [45]:

\[ HARM = 0.0148 \times HIC + 15.6 \times DAMAGE \]  

\[ HPS = \sum_{i=1}^{18} M_i(HARM)_i \]  

where the weighting factors, \( M_i \), were determined based on impact location and speed from the review of concussive NFL impacts [45]. This work has recently been adopted by the NFL for their assessment of football helmets [62], however not enough time has elapsed to assess whether this has resulted in a reduction of the incidence of concussion.

2.2.1 Limitations

Head injury models developed in conjunction with, or verified using FE models are subject to the limitations associated with the validation of those models. Recent studies by Zhou et al., and Zhao and Ji have highlighted major inconsistencies
in the strains calculated by displacement-validated FE models, suggesting that current methods for model validation which rely heavily on cadaveric displacement data are not effective for strain calculation [59,63,65].

Specific examples of these inconsistencies include work by Ji et al., which compared SIMon, the Dartmouth Scaled and Normalized Model (DSNM), and the Wayne State University Head Injury Model (WSUHIM) using characteristic head impacts frequently seen in ice hockey [59]. Their findings showed that the validated models differed significantly in their calculated strain and strain rate, with peak MPS differing by 75% for one impact [59]. Similarly, Giordano and Kleiven evaluated the THUMS, GHBMC, and KTH models against cadaveric pressure, displacements, and strains, which further illustrated significant differences in their performance for identical input parameters [63].

These studies suggest the necessity of additional validation data, specifically high and mid-rate cadaveric strain data, for FE models used for strain estimation, as validation against displacement data does not necessarily result in biofidelic strain data [65]. These limitations to FE models suggest that the existing severity
indices are also subject to this variability and would require recalibration as the FE models are adapted in response to additional validation data.

Given the current techniques available for helmet evaluation and the limitations associated with intracranial strain prediction from kinematic data and FE models, there is a need to advance new tools and methodologies capable of directly resolving the causes of concussion during impact testing. A deformable physical model, capable of resolving intracranial strain and calibrated to clinical outcomes of injury, would be invaluable to the advancement of helmet design and evaluation, ultimately leading to a reduction in the incidence and severity of mTBI.

2.3 Deformable Head Surrogates

As literature on deformable head surrogates for impact investigations is limited, the following section will include studies focused on the use of deformable head surrogates in the evaluation of blast-induced TBI. The shock wave physics field has previously developed a number of deformable head surrogates for the investigation of blast-induced TBI, which have been broadly limited to measurements of cranial pressure and head kinematics.

Early headforms investigating blast-induced TBI consisted of simple spherical and egg-shaped closed-head physical models subjected to blast waves. For example, a model created by Zhu et al. consisted of a polyethylene shell filled with Sylgard 527 and a central pressure transducer recording the internal pressure time history (Figure 6a) [66]. Another study by Alley et al. filled spherical polymethyl methacrylate (PMMA) shells with synthetic gels (either Perma-Gel or polydimethylsiloxane (PDMS)), and recorded internal pressure at 3 locations and external acceleration [67]. Alley et al. also created a thin slice representation of their spherical model, consisting of a PMMA ring cast with Perma-Gel, and painted
with a speckle pattern [67] for the measurement of strain and displacement using digital image correlation (DIC) (Figure 6b) [67].

Figure 6: Spherical surrogates for the modelling of blast-induced TBI. (a) Egg-shaped surrogate shown with inserted optic pressure gauge. Reproduced from Zhu et al. [66]. (b) Strain field in Perma-Gel at maximum strain for 8 in. standoff distance [67]. Figure and caption reproduced from Alley et al. [67].

Other studies made significant strides towards the creation of higher biofidelity headforms for the investigation of blast-induced TBI. The headform created by Awad et al. [68] (Figure 7a) consisted of a human-modelled polyurethane resin skull, a Sylgard-527 brain cast with gyrification of a similar size and shape to that of the human brain, and rubber skin surrogate [68]. This headform measured intracranial pressure, as well as the acceleration of the head surrogate and the brain surrogate, directly [68]. The headform developed by Defense Research and Development Canada (DRDC), the Brain Injury Protection Evaluation Device (BIPED), included a polymeric skull, a Sylgard 527 brain with a cast cerebellum, skin surrogate, falx and tentorium membrane simulants, and was perfused with saline to simulate cerebrospinal fluid (CSF) [29]. The headform measured external pressure (Figure 7b), intracranial pressure, and acceleration of the headform in response to
blast loading [29]. Intracranial pressure measurements made by the BIPED headform have been validated against cadaveric data [28]. A study by Li et al. has also found the acceleration time-histories of the BIPED headform in response to impact to be biofidelic [69]. While these headforms have marked significant advancements in the inclusion of anatomical simulating structures, their measurement capabilities continue to focus primarily on pressure and kinematics.

Figure 7: Headforms for the evaluation of blast-induced TBI by Awad et al. [68], and Ouellet and Philippens [29]. (a) Brain surrogate shown inside the skull surrogate with mounted pressure transducers and accelerometer. Reproduced from Awad et al. [68]. (b) BIPED skull shown with external pressure transducers [29]. Reproduced from Ouellet and Philippens [29].

2.3.1 Deformable Headforms for Modeling Impact Events

Two recent studies have marked advancements in the development of deformable head surrogates for impact loading. A study by Petrone et al. constructed a headform for the evaluation of helmets from MRI scans of a 50th percentile male human
The headform, shown in Figure 8, consisted of a 3D printed acrylonitrile butadiene styrene (ABS) skull, a silicone rubber brain, silicon skin, and was filled with silicon oil to simulate cerebrospinal fluid. The headform skull contained 10 pressure sensors, and the brain surrogate contained 9 accelerometers, placed at the exterior of the lobes, in the cerebellum, and in the center of mass of the brain. The brain surrogate also housed a three-axis gyroscope, placed at its center of mass [70]. The headform was subjected to pendulum impacts at the back, side and front of the head surrogate [70]. This headform marked significant efforts to improve the biofidelity of helmet-evaluating head surrogates, and provided additional measurement capabilities when compared to the industry standard headforms [32, 39, 48]. However, parameters resolved in this headform continued to be limited to pressure and acceleration, and did not resolve intracranial strain.

**Figure 8:** Headform by Petrone et al. [70], helmeted and aligned for a rear impact. Reproduced from Petrone et al. [70].

Taking a different approach, a study by Singh et al. appeared to borrow from earlier blast-focused surrogates [67], creating a simplified half-head model capable of measuring strain during impact testing (Figure 9) [71]. The headform included a silicone skin, a polylactic acid (PLA) skull, and a gelatin brain. The headform was
subjected to impacts and the strain fields in the optically-accessible brain were analyzed using DIC [71]. This work provided valuable insights into the effects of boundary conditions on strain fields within the brain surrogate. The inclusion of a PLA tentorium and falx illustrated strain concentrations in the regions of these additional boundary conditions [71]. However, the complex surface boundary created by the open model introduced a vulnerability of the brain surrogate to decoupling from the skull surrogate, and Rayleigh wave interference. The optical accessibility requirements of traditional DIC restricted the addition of further boundary conditions to address these issues.

![Strain-measuring half-head surrogate by Singh et al. [71]. Reproduced from Singh et al. [71].](image)

While a concerted effort has been made to improve the complexity and biofidelity of headforms, these models continue to be limited in their measurement capabilities. As evidence has linked intracranial strain to mTBI and concussion [9,25], there remains a need to bridge the gap between simple surrogates capable of making in situ strain measurements [71] and higher biofidelity headforms that better represent the anatomical features of the human head and are capable of
wearing helmets for evaluation [29, 68, 70].

2.4 Strain Analysis Techniques

A variety of methodologies exist for the measurement of strain [72, 73]. However, many of these techniques require maintained physical contact with the material being measured [72, 73]. As the goal of these headforms is to provide a response that is adequately representative of brain tissue strain, the material under investigation would likely be a highly deformable elastomer. As such, it is of paramount importance that the tools used for measurement offer minimal interference in the material response. Additionally, given the nature of the timescale of head impact events (initial loading \( \sim 20 \) ms, complete response \( < 100 \) ms), the measurement rate must be of adequate temporal resolution. For this reason, optical methods for strain measurement, such as digital image correlation (DIC), are of particular interest for this application.

DIC is a non-contact, optical technique for the measurement of displacement and strain within a specimen undergoing physical deformation [74, 75]. Randomized patterns of distinct pixels representing a physical surface on the sample are tracked in a sequence of digital images by obtaining a one-to-one mapping of pixel patterns between an initial, undeformed reference image and the subsequent images in the deformation sequence [74, 75].

However, the need for the head surrogate to be equipped with headgear for helmet testing would create an optical obstruction to the brain surrogate. Additionally, the work by Singh et al. [71] illustrated the need for more restrictive boundary conditions on the brain surrogate to prevent the propagation of Rayleigh waves.
To allow for helmet testing and more restrictive and biofidelic boundary conditions in a head surrogate, DIC techniques which permit the analysis of optically-inaccessible surfaces, such as X-ray DIC, may be favorable.

### 2.4.1 Digital Image Correlation Algorithms

This section reviews the algorithms used in Digital Image Correlation. In particular, this work focuses on the algorithms used in an open-source MATLAB-based DIC script, NCorr [76].

The DIC algorithm creates the mapping from the reference image to subsequent images by subdividing the region of interest (ROI) into subsections (subsets) in the reference image and determining their locations in each subsequent frame of the image sequence using a linear, first-order transformation [75]:

\[
\begin{align*}
x_{\text{cur}_i} &= x_{\text{ref}_i} + u_{\text{rc}} + \frac{\partial u}{\partial x_{\text{rc}}} (x_{\text{ref}_i} - x_{\text{ref}_c}) + \frac{\partial u}{\partial y_{\text{rc}}} (y_{\text{ref}_i} - y_{\text{ref}_c}) \\
y_{\text{cur}_j} &= y_{\text{ref}_j} + v_{\text{rc}} + \frac{\partial v}{\partial x_{\text{rc}}} (x_{\text{ref}_i} - x_{\text{ref}_c}) + \frac{\partial v}{\partial y_{\text{rc}}} (y_{\text{ref}_i} - y_{\text{ref}_c})
\end{align*}
\]

where \(x_{\text{ref}_i}\) and \(y_{\text{ref}_j}\) represent the coordinates of a point within the initial reference subset, \(x_{\text{ref}_c}\) and \(y_{\text{ref}_c}\) are the coordinates of the center of the reference subset, and \(x_{\text{cur}_i}\) and \(y_{\text{cur}_j}\) are the coordinates of the point within the current (deformed) subset [75].

The goal of the DIC algorithm is to determine the deformation vector, \(\vec{p}\), which provides an optimal match between coordinates \(x_{\text{ref}_i}\) and \(y_{\text{ref}_j}\) in the reference subset, and \(x_{\text{cur}_i}\) and \(y_{\text{cur}_j}\) in the current subset, based on correlation criteria [75]. DIC typically uses one of two forms of correlation criteria to quantify the similarity between grayscale pixel values within a reference subset and a current subset: a cross-correlation criterion, or a normalized least squares criterion. In the case
of the open-source MATLAB-based DIC script, NCorr [76], the algorithm uses a forward-additive Gauss-Newton iterative optimization approach to minimize the least squares correlation criterion between the reference and current subset (Figure 10) [75].

**Figure 10:** Graphical representation of the forward-additive Gauss-Newton process for aligning current data to reference data. Subset data is represented by solid black lines, dotted red lines represent incremental changes to the deformation vector [75].

Displacement data can then be calculated for points located at the center of initial reference subsets, from which a displacement data grid is generated within the ROI [75]. The displacements are then reduced or interpolated to form a full displacement field for the region of interest [75]. To reduce the need for interpolation, it is beneficial to ensure subset spacing is defined such that overlap exists between subsets.

From the displacement field, the Green-Lagrangian strain is calculated, given by the following strain tensor [75]:

\[
E = \begin{bmatrix}
E_{xx} & E_{xy} \\
E_{xy} & E_{yy}
\end{bmatrix}
\]  

(13)

The components of the tensor are described in equations [14]–[16] where \( u \) and \( v \) represent orthogonal displacement data in the local coordinate system [75].
\[ E_{xx} = \frac{1}{2} \left( 2 \frac{\partial u}{\partial x} + \left( \frac{\partial u}{\partial x} \right)^2 + \left( \frac{\partial v}{\partial x} \right)^2 \right) \] (14)

\[ E_{xy} = \frac{1}{2} \left( \frac{\partial u}{\partial y} + \frac{\partial v}{\partial y} + \frac{\partial u}{\partial x} \frac{\partial u}{\partial y} + \frac{\partial v}{\partial x} \frac{\partial v}{\partial y} \right) \] (15)

\[ E_{yy} = \frac{1}{2} \left( 2 \frac{\partial v}{\partial y} + \left( \frac{\partial u}{\partial y} \right)^2 + \left( \frac{\partial v}{\partial y} \right)^2 \right) \] (16)

To determine \( E_{xx}, E_{xy}, \) and \( E_{yy}, \) an overconstrained system of equations is developed using a least squares plane fit on subsets of the displacement data [75]:

\[
\begin{bmatrix}
1 & x_{ref_{first}} - x_{c} & y_{ref_{first}} - y_{c} \\
\vdots & \vdots & \vdots \\
1 & x_{ref_{last}} - x_{c} & y_{ref_{last}} - y_{c}
\end{bmatrix}
\begin{bmatrix}
a_{u,\text{plane}} \\
\frac{\partial u}{\partial x_{\text{plane}}} \\
\frac{\partial u}{\partial y_{\text{plane}}}
\end{bmatrix}
= \begin{bmatrix}
u_{rc}(x_{ref_{first}} , y_{ref_{first}}) \\
\vdots \\
u_{rc}(x_{ref_{last}} , y_{ref_{last}})
\end{bmatrix}
\] (17)

where \( u_{\text{plane}} \) and \( v_{\text{plane}} \) are given by:

\[ u_{\text{plane}}(x, y) = a_{u,\text{plane}} + \left( \frac{\partial u}{\partial x_{\text{plane}}} \right)x + \left( \frac{\partial u}{\partial y_{\text{plane}}} \right)y \] (18)

\[ v_{\text{plane}}(x, y) = a_{v,\text{plane}} + \left( \frac{\partial v}{\partial x_{\text{plane}}} \right)x + \left( \frac{\partial v}{\partial y_{\text{plane}}} \right)y \] (19)

Using the algorithm and techniques described above, full-field displacement and strain data can be extracted from a sequence of digital images depicting a surface undergoing deformation. In traditional DIC applications, this requires the deformed surface to be optically accessible for imaging.
2.4.1.1 X-Ray Digital Image Correlation

DIC may also be performed on X-ray images, using the same methodologies described above, allowing for the analysis of otherwise optically-inaccessible surfaces. In the case of X-ray digital image correlation (XDIC), rather than using an optical pixel pattern on an accessible surface, such as paint speckles or surface roughness maps, the pixel pattern is generated using varying contrast levels within the X-ray images that occur due to density variations within the materials. In the case of a physical brain surrogate, the elastomeric material used to create the surrogate presents with uniform contrast within the X-ray images. Therefore to perform XDIC, radiopaque contrasting agents must be added to the brain surrogate to create varying levels of attenuation, creating a pixel pattern in the radiographs, as was done by Synerggren et al., who inserted lead grains into polyester samples to measure the internal deformation in response to an impact by a 9 mm steel ball bearing (Figure 11) [77].

2.4.2 Limitations of Digital Image Correlation

Digital image correlation is typically found to be most suitable for quasi-static loading conditions, or loading conditions where bulk-body motion is minimal. As such, to successfully apply DIC to highly dynamic loading, such as a head-impact loading scenario, motion of the ROI must be isolated from the bulk motion of the head surrogate.

To separate motion in a region of interest from the bulk rigid-body motion, an image stabilization script was written in MATLAB by Dutrisac [78]. The algorithm tracks user-selected, rigidly attached fiducial markers through the entire image sequence and computes a transform mapping the initial position of the fiducial
Figure 11: X-ray image sequence showing the impact of a 9 mm diameter ball bearing into a polyester specimen with embedded lead grains, at 5μs intervals. Reproduced from Synnergren et al. [77].

Markers to their positions in each subsequent frame. Images are translated, rotated, and scaled such that the fiducial markers in each frame match the positions of the initial reference frame fiducial markers. The result is a change of frame of reference from Eulerian to Lagrangian, allowing for clear observation of deformation within the ROI. An overview of the image stabilization process can be seen in Figure 12.

Additionally, the use of DIC with two-dimensional images relies on the assumption that any out-of-plane displacement and strain is minimal, as these deformations would not be captured by the planar images. In the case of a headform undergoing impact loading, care must be taken to ensure experiments are aligned
such that out-of-plane motion of the physical head surrogate is reduced. Additionally, an investigation should be performed to verify the validity of these assumptions within the brain surrogate.

2.5 Summary

To date, insights gained from the use of headforms in helmet testing have been limited to analyses of rigid-body head kinematics [17,18,32]. Kinematic data produced by these tests are used in conjunction with various kinematic and computational models of injury to link head impact tests to the mechanisms of concussion [45,49,50,61]. However, recent studies have highlighted significant inconsistencies between the strain predictions of validated models [59,63,64], indicating the need for additional validation data and more robust model validation protocols [65]. Alternatively, the development of physical head models capable of resolving the causes of brain injury would allow for the establishment of a more
direct link between testing conditions and clinical outcomes of injury.

An investigation of deformable head surrogates used in the shock-wave physics field for the investigation of blast-induced TBI has shown promising advancements in the development of higher-biofidelity, deformable headforms [29, 68]. However, these headforms have been primarily limited to measurements of intracranial pressure and head kinematics. Recent work by Singh et al. demonstrated the use of a half-head surrogate capable of measuring in-situ intracranial strain using DIC [71]. However, optical-accessibility requirements make this an unsuitable tool for helmet testing. Work by Synnergren et al. has illustrated the ability to measure internal deformation using X-ray DIC with the insertion of radiopaque materials, embedded in the sample under deformation [77]. This technique is of significant interest in the progression of deformable headforms under impact loading, as the brain surrogate would be optically inaccessible in a closed-skull model, necessary for helmet testing.

The following chapters of this thesis will present a novel head surrogate for the evaluation of mTBI in response to head impacts, with a focus on helmet evaluation. This headform, an adaptation of the BIPED headform [29], is based on measuring intracranial strain using high-speed X-ray digital image correlation. Efforts have been made to validate this iteration of the BIPED headform against cadaveric impact data, and to provide insights to further guide the design and development of future iterations of the BIPED headform.
Chapter 3

Methodology

3.1 BIPED Headform

The BIPED headform is a head surrogate designed and manufactured by Defence Research and Development Canada Valcartier, which has been previously validated under blast loading conditions \cite{29}. The headform, seen in Figure \ref{fig:headform}, features multiple anatomy simulating structures including skin, skull, brain, cerebellum, artificial cerebrospinal fluid, falx, and tentorium (not included in this iteration).

To enable displacement field measurements during impact testing, the current iteration of the BIPED headform has been modified during the manufacturing process to include the insertion of radiopaque markers. The basis for the marker design was a prior study that sought to optimize design parameters for minimal interference of embedded markers on the mechanical response of a tissue-simulating gel \cite{79}. In accordance with the results of this study, the markers used for the BIPED headform consisted of a thermoplastic gel (Humimic gel #4) containing 60% wt. Barium Sulphate (BaSO$_4$) powder. This design configuration optimized both marker response in the tissue-simulating gel and radiopacity for visibility under high-speed X-ray imaging. The cylindrical markers had a 2 mm diameter and
length, a density of 1.6 g/cc, similar to nominally neutral density markers used in prior cadaveric research \cite{80,81}, and a nominal elastic modulus of 12 kPa \cite{79}.

\begin{figure}[h]
\centering
\includegraphics[width=\textwidth]{image}
\caption{(a) Photograph of the disassembled BIPED headform, (b) markers within the brain surrogate during casting process, (c) high resolution X-ray of the headform in the sagittal orientation.}
\end{figure}

During the manufacturing process, markers were inserted in the Sylgard 527 brain using a three-part mould for the casting procedure. The first section of the mould was filled and allowed to fully cure at room temperature, extending 12 mm past the sagittal plane. The second section of the mould, a 3 mm thick layer, was filled with Sylgard 527 and allowed to partially cure at room temperature for 3 hours. Approximately 500 elastomeric markers were then randomly distributed over the partially cured layer of Sylgard 527 to ensure that the markers penetrated in a controllable manner, limiting out-of-plane marker drift (Figure 13b). The marker layer was allowed to fully cure at room temperature, after which the final section of the brain surrogate was cast in the mould over the existing surface and allowed to fully cure under vacuum at room temperature. A sagittal view of the marker plane within the BIPED headform can be seen in the high-resolution X-ray image provided in Figure 13c.
3.2 Impact Testing

A number of impact testing experiments were performed to evaluate the intracranial strain response within the elastomeric brain surrogate, using a pneumatically-driven linear impactor (Cadex Inc.) (Figure 14a) and a twin-wire drop tower (Figure 14b). In each of these impact experiments, the headform was mounted to the impact platform using an unbiased neckform, a variation of the Hybrid III neck that permitted flexion in all directions [82, 83]. The neckform was tensioned by applying a 1.13 Nm torque to the retaining nut prior to the impacts, and was subsequently verified and adjusted, as necessary.

Figure 14: (a) Pneumatically-driven linear impactor, and (b) Twin wire drop tower, reproduced from Cadex Inc. [84].

During testing, the headform was perfused with water at a static pressure of
1,000 mm of H$_2$O, which is within the range of normal intracranial pressures measured in adults [85]. This provides a slip boundary condition between the brain and skull surrogates, allowing for brain-skull relative motion as was observed in cadaveric specimens [86]. Perfusion was verified visually through the semi-transparent surrogate skull and X-ray imaging. A diagram of the perfusion system used in the impact experiments can be seen in Figure 15. Impact conditions varied based on experiment, and are summarized in Table 1.

![Diagram of perfusion system](image)

**Figure 15:** System used for headform perfusion.

### 3.2.1 Helmet Impact Testing

To evaluate the ability of this testing methodology to provide insights on the influence of helmets on intracranial strain, a series of linear impacts was performed using the pneumatically-driven linear impactor. The linear impactor was equipped
with a 13 kg ram and a high compliance 85 mm vinyl nitrile (VN) foam impactor face. The density of the VN foam was 110 kg/m$^3$. The headform was mounted to the linear bearing platform of the impactor via the unbiased neckform (Figure 16), however the platform was pinned to prevent sliding upon impact.

The impacts were performed in two headform orientations, frontal and rear, and each performed with a bare headform and a helmet-equipped headform (commercially available ice hockey helmet). The helmet selected for this study was not of significant importance, as the goal of performing helmeted impacts was to demonstrate the testing methodology with any commercially available helmet. For the frontal impact orientation, the headform was angled approximately 15° forward to ensure contact with the ‘frontal bone’ region of the headform skull, and impact speeds ranged from 3 m/s to 5 m/s. For the rear impact orientation the headform was upright and impacted at ram speeds ranging from 2.5 m/s to 3.5 m/s. Impact conditions were chosen to simulate a range of reasonably aggressive impacts that did not risk fracturing the headform in the bare impact configuration.

### 3.2.2 Repeatability Impact Testing

An additional series of impacts was performed to analyze the variability measured in the results of repeated impacts, and to verify the appropriateness of reporting results in terms of a representative peak percentile value. For this investigation, 10 bare frontal impacts were performed at an impact speed of approximately 4 m/s using the experimental setup described in Section 3.2.1.

### 3.2.3 Error Quantification Testing

To verify the validity of the assumptions that any out-of-plane response in the brain surrogate is minimal, an investigation was performed using the twin-wire
Figure 16: Photographs of the BIPED headform aligned on the impactor table in the (a) bare and (b) helmeted frontal impact orientations.

drop tower. For this experiment, the BIPED headform was aligned under a twin-wire drop tower, mounted to a rigidly attached table via the unbiased neckform \[82,83\]. A mass was dropped onto the BIPED headform while imaging the coronal plane of the head surrogate (Figure 17). The dropped mass consisted of a drop carriage and a mounted impactor cap comprising a 26.5 mm nylon backing, 35.7 mm of VN foam midlayer, and rounded nylon contact face (overall thickness 20.0 mm, 60.7 mm radius), with a total mass of 865 g. The BIPED was angled 45° forward to ensure that the dropped mass would contact the superior-posterior parietal bone region of the BIPED skull. Three drops were performed with drop heights that ranged between 125 mm and 130 mm, and impact speeds that spanned approximately 3.4 m/s to 3.7 m/s.
3.2.4 Impact Testing for Validation Against Cadaveric Data

To validate the BIPED data, a series of impacts was performed to ensure a direct comparison to existing cadaveric data collected using an identical impact platform [87]. For these impacts, the BIPED headform was mounted to the pinned linear bearing platform of the impactor via the unbiased neckform [82,83], and angled 15° forward to ensure impactor contact with the frontal bone region of the headform. The impactor was equipped with a 13 kg impact ram and a low-compliance, 50.8 mm thick neoprene impactor face. Three frontal impacts were performed with impact ram speeds ranging from 1 m/s to 2.1 m/s, carefully matched to the lower impact speeds of specimen #CO-103 from Dutrisac et al. (2021) [87]. These impact speeds were selected to minimize the risk of fracture to the BIPED headform.

Figure 17: BIPED headform aligned under twin-wire for coronal-view drop impacts.
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**Table 1:** Summary of impact experiments.
3.3 Data Capture and Processing

3.3.1 X-Ray Imaging and Kinematic Data Collection

Impacts were carried out in the Carleton University Impact Research Lab, which features a linear impactor coupled with a custom-designed high-speed X-ray (HSXR) cineradiography system. All testing configurations described in this work made use of the same data collection methodologies. Impacts were imaged using the HSXR system a frame rate of 5,000 fps. This frame rate was selected to ensure a high temporal resolution, while balancing visual clarity of the marker plane and the total time-scale that could be analyzed given limitations to computational processing power. The acceleration of the BIPED headform was recorded using a DTS 6DX Pro accelerometer (DTS Inc, USA), rigidly mounted to the base of the skull and translated to the center of gravity of the headform. Acceleration data was captured at 20 MHz and downsampled to 20 kHz, after which it was filtered using a low pass CFC180 filter at 300 Hz \[88\] to eliminate high-frequency noise.

3.3.2 Image Processing

Image processing was performed to improve marker visibility within the headform. Images were first corrected for distortion (Figure 18), after which adjustments were made for contrast and flat-field to ensure marker plane visibility throughout the entire image sequence (Figure 19). Contrast and flat-field adjustments were made using the Image Processing Toolbox in MATLAB \[89\].

It was found that the most successful approach to extracting displacement and strains from image sequences involved stabilizing the image sequence prior to analyzing the relative motion of the brain. This was performed using an in-house
Figure 18: Image distortion calibration grid (a) before and (b) after image correction. Certain portions of the image have been highlighted yellow to emphasize changes made during image correction.

Figure 19: Photographs of the BIPED headform (a) before and (b) after flat field and contrast adjustments.

MATLAB algorithm [78] to fix the image around a feature on the headform and isolate the relative motion within the head from its bulk motion. A sample stabilized image sequence can be seen at 10 ms intervals in Figure 20.

3.3.3 X-ray Digital Image Correlation

Digital image correlation (DIC) was used to extract the displacement and strain data from the stabilized image sequences. The stabilized images were imported into an open-source MATLAB-based DIC script, NCorr [76]. The region of interest
Figure 20: Sample stabilized image sequence, showing a 4 m/s bare frontal impact at 10 ms intervals.

(ROI) was defined using a binary png image which highlighted the marker plane region of the BIPED headform (Figure 21). The parameters used in the DIC calculation are shown in Table 2. Subset radius and strain radius parameters were selected to ensure detailed measurements of the displacement and strains within the marker plane, while reducing noise introduced by slight variations in the contrast within the X-ray images. Subset spacing distance was selected to maximize the overlap between subsets, and minimize the need for interpolation of displacement results.

Figure 21: Region of interest in the bare and helmeted headform. (a) X-ray image of the bare headform with highlighted ROI, (b) binary ROI image for bare headform impacts, (c) X-ray image of the helmeted headform with highlighted ROI, and (d) binary ROI image for helmeted headform impacts.
Strains extracted from the DIC calculation in this study are Green-Lagrangian. Strains presented as a percentile are based on approximately 38,000 pixels, accounting for 13% of the original image, and have been filtered using a 2\textsuperscript{nd}-order Butterworth filter with a 300 Hz cut-off frequency. This filter was found to eliminate high-frequency noise while maintaining a clear representation of the signal features. It should be noted that this data is not representative of the response of a single pixel location within the brain; rather it is representative of the percentile value, located at any position within the brain surrogate, at a given time.
Chapter 4

Results

4.1 BIPED Impact Study Results

A series of impacts was performed on the bare and helmeted BIPED headform to measure the response of the surrogate head and brain to impact. Impacts were performed in a frontal and rear impact configuration, using a high compliance foam impactor face to simulate a range of aggressive, subconcussive impacts, while ensuring a low risk of headform fracture in the bare configuration. A total of 5 bare and 5 helmeted frontal impacts, and 3 bare and 3 helmeted rear impacts were collected. Table 3 provides summary data for the head impacts, including kinematic data referenced to the CG of the headform, as measured using a 6DX PRO accelerometer package. Sample acceleration profiles can be seen in Figure 22.
<table>
<thead>
<tr>
<th>Impact Config.</th>
<th>Impact Speed (m/s)</th>
<th>Bare Headform $a_p$ (g)</th>
<th>$\alpha_p$ (rad/s$^2$)</th>
<th>$\omega_p$ (rad/s)</th>
<th>Helmeted Headform $a_p$ (g)</th>
<th>$\alpha_p$ (rad/s$^2$)</th>
<th>$\omega_p$ (rad/s)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Frontal</td>
<td>3.1</td>
<td>45.3</td>
<td>2085</td>
<td>17.9</td>
<td>38.3</td>
<td>2298</td>
<td>16.5</td>
</tr>
<tr>
<td></td>
<td>3.5</td>
<td>46.6</td>
<td>1842</td>
<td>20.6</td>
<td>37.4</td>
<td>2843</td>
<td>17.5</td>
</tr>
<tr>
<td></td>
<td>4.0</td>
<td>54.5</td>
<td>2302</td>
<td>25.4</td>
<td>43.1</td>
<td>2966</td>
<td>21.0</td>
</tr>
<tr>
<td></td>
<td>4.5</td>
<td>63.5</td>
<td>3124</td>
<td>26.8</td>
<td>43.4</td>
<td>2925</td>
<td>23.2</td>
</tr>
<tr>
<td></td>
<td>5.0</td>
<td>63.1</td>
<td>2841</td>
<td>29.9</td>
<td>52.1</td>
<td>3319</td>
<td>25.4</td>
</tr>
<tr>
<td>Rear</td>
<td>2.4</td>
<td>33.1</td>
<td>1540</td>
<td>16.2</td>
<td>24.9</td>
<td>1355</td>
<td>14.3</td>
</tr>
<tr>
<td></td>
<td>3.1</td>
<td>38.8</td>
<td>1760</td>
<td>21.1</td>
<td>34.3</td>
<td>1998</td>
<td>17.6</td>
</tr>
<tr>
<td></td>
<td>3.4</td>
<td>45.7</td>
<td>1934</td>
<td>23.4</td>
<td>38.1</td>
<td>2064</td>
<td>21.2</td>
</tr>
</tbody>
</table>

Table 3: Summary of the impact kinematics for the headform tests. Kinematic data was filtered at a frequency of 300 Hz in accordance with CFC180 [88]. Kinematic data has a linearity error of $< 1\%$ FS [90].

### 4.1.1 Brain Surrogate Displacements

The in-plane brain surrogate displacements were computed using XDIC. A high-resolution two-dimensional displacement field was generated for analysis by transposing the brain surrogate motion onto surrounding pixel locations. Sample full-field displacements can be seen in Figure 23 for both a frontal and rear 3.1 m/s bare impact.
Figure 22: Resultant linear acceleration profiles for (a) 3.5 m/s frontal impacts and (b) 3.4 m/s rear impacts; and resultant rotational acceleration profiles for (c) 3.5 m/s frontal impacts and (d) 3.4 m/s rear impacts. Kinematic data has been filtered at a frequency of 300 Hz in accordance with CFC180 [88]. Kinematic data has a linearity error of $< 1\%$ FS [90].

Of particular interest is the difference in the geometry of the displacement fields between the two impact orientations. In both impacts, we observe a clear rotation of the marker plane about the CG of the surrogate brain. This is consistent with observations made in Post-Mortem Human Surrogate (PMHS) impacts using a similar experimental configuration [87]. However, in the rear impact configuration, there appears to be significantly more vertical displacement in the superior
portion of the brain when compared to the frontal impact configuration, as well as a more elongated displacement path shape. This appears to agree with the displacement fields observed in impact C408-T4 by Hardy et al. [81]. This change in displacement field geometry may be related to the impact location, as frontal impacts were oriented 15° forward, while rear impacts were performed in an upright position, resulting in a more inferior impact location on the skull surrogate.

![Figure 23](image)

**Figure 23:** Brain surrogate marker displacement fields for 3.1 m/s bare impacts: (a) frontal, and (b) rear. The effects of impact orientation can be seen in the different geometries of the displacement traces.

We can also make comparisons between bare and helmeted impacts at similar impact speeds (Figure 24), and at similar peak rotational accelerations (Figure 25). At similar impact speeds, we observe that the displacement fields look very similar between bare and helmeted impacts, with slightly more vertical displacement observed in the superior frontal region of the helmeted impact. At similar peak rotational accelerations, we observe more horizontal displacement in the posterior region of the brain surrogate, and less vertical displacement in the frontal region of the brain surrogate for the helmeted impact.
Figure 24: Brain surrogate marker displacement fields for 3.5 m/s frontal impacts: (a) bare, and (b) helmeted.

Figure 25: Brain surrogate marker displacement fields for a (a) 3124 rad/s$^2$ bare frontal impact, and a (b) 2966 rad/s$^2$ helmeted frontal impact.
4.1.2 Maximum Principal Strain

From the internal displacement field measurements, the time-resolved strain distributions within the brain surrogate were determined. Figure 26 shows composite image sequences of the full-field maximum principal strain (MPS) distributions superimposed over HSXR images, allowing for the observation of regions of high strain within the headform throughout the entire impact event, at 5 ms intervals. Comparing a 3.1 m/s bare frontal impact (Figure 26a), and a 3.1 m/s bare rear impact (Figure 26b), we observe that while the timing of peak strains differ in the two impact orientations (45 ms in the frontal impact, and 20 ms in the rear impact), the regions of peak strain are similar.

**Figure 26:** MPS response in the headform surrogate for 3.1 m/s bare impacts: (top) frontal, and (bottom) rear, shown at 5 ms intervals. Peak strain can be observed at $\approx 45$ ms in the frontal impact and at $\approx 20$ ms in the rear impact.
The time-resolved MPS response within the brain surrogate was also plotted, shown in Figure 27a for a 3.1 m/s bare frontal impact. The data was further subdivided into regions that are representative of anatomical brain regions (frontal, parietal, occipital, insular, and cerebellum) to better visualize regional responses within the surrogate, as shown in Figure 27b through 27f. In these plots, each curve represents the MPS response at an individual pixel location within the brain surrogate. To manage the volume of data generated by the DIC calculations in the region of interest, these curves have been generated for a grid of pixels within the surrogate brain with an increased internode spacing of approximately 8.4 mm, rather than every pixel within the region of interest.

Figure 27: (a) MPS response within the headform for a 3.1 m/s bare frontal impact over 70 ms, and subdivided into the (b) frontal, (c) parietal, (d) occipital, (e) insular, and (f) cerebellum areas.

By subdividing the MPS response into representative anatomical regions, it was observed that certain regions, such as the occipital and cerebellum regions, displayed distinct and consistent strain responses. However, larger regions exhibited more broad strain distributions, such as the frontal region. Similar observations were made by Dutrisac et al. [87] on tissue displacements in the frontal lobe of a
PMHS brain, which spanned a much broader range than other lobes in the brain for the same impacts.

### 4.1.3 Shear Strain

Time-resolved plots were also generated to observe the shear strain response of the BIPED brain to impact, illustrated in Figure 28 for a bare and helmeted 3.1 m/s frontal impact. It was observed that the shear strain responses were very similar for both impacts despite the inclusion of a helmet, with the exception of the insular region in the brain surrogate. The differences observed in the insular region (Figure 28d, and 28i) may be attributed to a decreased region of interest in the helmeted impact due to interference from helmet hardware.

Peak strain rates were also calculated for the first and second phases of head surrogate motion (Table 4). The greatest shear strain loading rates were observed at approximately 35 ms following impact, during the inversion of the shear strain which occurred between the two phases of head surrogate motion.
<table>
<thead>
<tr>
<th>Region</th>
<th>Initial Shear Strain Loading Rate (s⁻¹)</th>
<th>Shear Strain Inversion Rate (s⁻¹)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Bare Headform</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Frontal</td>
<td>5.16</td>
<td>7.88</td>
</tr>
<tr>
<td>Parietal</td>
<td>5.00</td>
<td>5.80</td>
</tr>
<tr>
<td>Occipital</td>
<td>5.29</td>
<td>6.02</td>
</tr>
<tr>
<td>Insular</td>
<td>3.97</td>
<td>8.14</td>
</tr>
<tr>
<td>Cerebellum</td>
<td>3.44</td>
<td>3.44</td>
</tr>
<tr>
<td>Helmeted Headform</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Frontal</td>
<td>6.12</td>
<td>9.40</td>
</tr>
<tr>
<td>Parietal</td>
<td>5.45</td>
<td>7.46</td>
</tr>
<tr>
<td>Occipital</td>
<td>6.98</td>
<td>7.98</td>
</tr>
<tr>
<td>Insular</td>
<td>4.58</td>
<td>9.75</td>
</tr>
<tr>
<td>Cerebellum</td>
<td>5.23</td>
<td>9.61</td>
</tr>
</tbody>
</table>

*Table 4:* Summary of the shear strain rates, subdivided by brain surrogate region, for a 3.1 m/s frontal impact.
Figure 28: Headform shear strain response over 70 ms for a 3.1 m/s frontal impact in the bare: (a) frontal, (b) parietal, (c) occipital, (d) insular, and (e) cerebellum regions; and helmeted: (f) frontal, (g) parietal, (h) occipital, (d) insular, and (e) cerebellum regions. High-rate strain inversions can be seen at $\approx 35$ ms.
4.2 Repeatability

To analyze the variability in the results measured using the BIPED headform, a repeatability study was conducted by performing a series of 10 bare frontal impacts at an impact speed of approximately 4 m/s. A summary of the kinematic data from this test series is presented in Table 5. Figure 29 illustrates the high degree of repeatability that was observed in the displacement fields produced by repeated headform impacts on a singular headform.

<table>
<thead>
<tr>
<th>Impact Speed (m/s)</th>
<th>$a^p$ (g)</th>
<th>$\alpha^p$ (rad/s$^2$)</th>
<th>$\omega^p$ (rad/s)</th>
</tr>
</thead>
<tbody>
<tr>
<td>4.04 ± 0.02</td>
<td>52.4 ± 1.3</td>
<td>2391 ± 95</td>
<td>25.5 ± 0.2</td>
</tr>
</tbody>
</table>

Table 5: Summary of the impact kinematics for the repeatability impact tests. Mean values are presented with corresponding standard deviations. Kinematic data has a linearity error of < 1% FS [90].

In addition to verifying the repeatability of the measurements for a given headform, this study served to inform the appropriateness of summarizing results in terms of a single MPS value or other representative peak values of interest. Given the quantity of data produced by each XDIC calculation, as seen in Figures 27 and 28, the ability to summarize an impact event with representative results would be greatly beneficial for making direct comparisons between various impacts. Nevertheless, the use of peak values carries the risk of introducing errors caused by ill-resolved strain fields at the boundaries of the region of interest. Though the 95th percentile MPS is the most commonly used value in the computational head injury modelling community [91-94], it is unclear if this value would be appropriate for the current brain strain model without conducting a thorough investigation based on an objective measurement from the test series.

For this analysis, a statistical distribution of the MPS response from each pixel
in the ROI was generated for each of the 10 impacts. Peak MPS was calculated for the 50th percentile to the 100th percentile, for a timescale of 35 ms (prior to significant neckform engagement) and on the complete 70 ms timescale. The average percentile values for the ten impacts are presented, along with their standard deviations in Figure 30a. To better identify trends in scatter and noise among the dataset, the standard deviations were plotted against MPS percentile separately in Figure 30b. It was found that the standard deviations for MPS were consistent up to the 99th percentile peak values when considering the initial phase of motion.
(35 ms), and up to the 92\textsuperscript{nd} percentile peak values when considering the complete timescale (70 ms). The same analysis was conducted on the shear strain, seen in Figure 30c and Figure 30d. The standard deviations on the calculated shear strains were more consistent, with no clear point of deviation. While this analysis did not consider variability introduced into the measurements by the manufacturing processes used to create the BIPED headform, it was determined that the strain fields measured in a singular BIPED headform are highly repeatable.

**Figure 30:** Mean peak (a) MPS and (c) shear strain plotted from 55\textsuperscript{th} to 100\textsuperscript{th} percentile, with error bars representing standard deviation. Standard deviation plotted for 55\textsuperscript{th} to 100\textsuperscript{th} percentile (b) MPS and (d) shear strain. A sharp rate change in MPS standard deviation can be seen at the 92\textsuperscript{nd} percentile.
4.3 Error Quantification Results

A series of analyses was performed to quantify the error bounds on the measured results in the head surrogate.

4.3.1 Measurement Error

To investigate the measurement error introduced into the strain results by the measurement methodology, a series of static HSXR images was corrected for distortion and adjusted for visibility, stabilized, and analyzed using XDIC. It was found that the 16 ms image sequence had a scatter range of -0.63% to 1.53% MPS, and -1.26% to 0.77% shear strain, indicating a measurement error of \(< 2\%\) strain.

4.3.2 Error Caused by Out-of-Plane Motion

To verify this methodology’s assumptions with respect to out-of-plane motion, an analysis was performed on a series of drop impacts with the BIPED head surrogate oriented in a coronal view (Section 3.2.3). Table 6 summarizes the total out-of-plane displacement, within unstabilized 1280 x 960 image sequences.

<table>
<thead>
<tr>
<th>Mass Drop Height (mm)</th>
<th>Estimated Mass Impact Speed (m/s)</th>
<th>Total Out-of-Plane Displacement (px)</th>
<th>Total Out-of-Plane Displacement (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>125</td>
<td>3.4</td>
<td>10.4</td>
<td>4.5</td>
</tr>
<tr>
<td>127</td>
<td>3.6</td>
<td>7.3</td>
<td>3.2</td>
</tr>
<tr>
<td>128</td>
<td>3.7</td>
<td>6.0</td>
<td>2.6</td>
</tr>
</tbody>
</table>

Table 6: Summary of out-of-plane displacements resulting from drop impacts.

An analysis was conducted on the worst-case out-of-plane motion, 4.5 mm, to determine its effect on displacement data. Figure 31a illustrates this analysis (not to
scale), which considered the scatter that would be caused on the X-ray projections should a point be displaced 4.5 mm out-of-plane, in either direction, at both the center and edge of the X-ray beam. It was determined that the maximum scatter on the displacement data caused by this motion would be -1.07 mm to +1.06 mm.

![Figure 31: (a) Schematic of displacement measurement scatter caused by out-of-plane marker translation. (b) Marker grid and strain elements used for the calculation of perceived strain caused by out-of-plane marker translation.](image)

A second analysis was performed considering the perceived strain that would be caused by this displacement scatter. For this analysis, a 3x3 grid of points with 4 mm spacing, similar to the marker spacing in the BIPED headform, was considered at the center and edge of the X-ray beam. The grid was translated 4.5 mm out-of-plane in either direction, similarly to Figure 31b, and the relative displacement between the two projected grids were used to calculate Green-Lagrangian strain.
for the triangular elements shown in Figure [31b]. The scatter on the MPS caused by
the movement of the marker grid was determined to be -0.56% to +0.57% strain.
Chapter 5

Discussion

5.1 Technique Advantages

This technique for helmet evaluation allows for a departure from the traditional focus of helmet testing and design, which has sought to assess and change the kinematic response of the head to impact through the reduction of peak linear and rotational accelerations. To date, the assessment of intracranial strain during head impacts has been limited to numeric and computational modelling estimates of MPS, calculated from kinematic input data [46, 91, 92, 95, 96]. By contrast, this headform allows for the simultaneous measurement of kinematic response, as well as measurements of time-resolved, full-field two-dimensional intracranial strain fields. The ability to observe the progression of the strain field throughout the impact event presents a unique opportunity for an improved understanding of the interactions between the various components of protective equipment and the brain surrogate, allowing for a clearer link between impact event, tissue strain, and injury.

For instance, analyzing the shear strain distributions shown in Figure 28 allowed for the observation of a high-rate strain inversion, which occurred at a rate higher than the initial shear strain loading rate (Table 4). This was observed in both
the bare and helmeted headform impacts in this study, suggesting that if injurious, this phenomenon continues to be of concern despite helmet use. The strain inversion may be explained by the phases of relative motion between the brain and skull surrogates. Upon impact, there is an initial phase of motion where the brain surrogate moves towards the site of impact, causing a forward rotation of the brain surrogate relative to the skull, peaking at approximately 35 ms post-impact. Following this phase, there is a counter-rotation of the brain surrogate relative to the skull as it moves towards the opposite side of the impact, rotating about its center of gravity. The shear strain inversion is therefore caused by the completion of the initial rotation and the subsequent counter-rotation of the brain relative to the skull. The strain inversion also coincided with the neckform stopping the initial motion of the headform and returning it to its neutral position (Figure 32).

**Figure 32:** Headform positioning at (a) the moment of impact, (b) peak neck flexion, and (c) returning to neutral position.

While the neckform used in this work would have influenced both the rate and peak of the shear strain and the shear strain inversion, it should be noted that a human neck would serve the same purpose of restricting the motion of the head in response to an impact, causing similar loading patterns in the brain. These loading patterns were also observed in the displacements fields measured in cadaveric impacts using the same experimental setup [87].
Figure 33: (a) Peak linear acceleration, (b) peak rotational speed, and (c) peak rotational acceleration data, plotted against impact ram speed.
To allow for a broader comparison of the data generated from the impacts presented in Section 3.2.1, peak kinematic values were plotted against impact speed (Figure 33), and the 92\textsuperscript{nd} percentile MPS and the 95\textsuperscript{th} percentile shear strain time histories were calculated and are presented in Figure 34 for the complete testing series. The 92\textsuperscript{nd} percentile MPS was selected as it was found to be the most reproducible MPS value for this headform in the repeatability study (Section 4.2). The 95\textsuperscript{th} percentile was selected for the shear strain as it was found to be as reproducible as other percentiles, while still eliminating any noise that may be present at the 100\textsuperscript{th} percentile due to ill-resolved strains at the ROI boundaries. Additionally, peak MPS were also plotted against peak kinematic indicators (Figure 35) to allow for observations on the correlation between kinematic indicators and MPS.

From the kinematic data, it was observed that helmeted impacts resulted in a consistent reduction in peak linear accelerations and peak rotational rates when compared to similar impact speeds on the bare headform, in both the frontal and rear impact configurations (Figure 33a, and 33b). However, this trend was not observed in the peak rotational accelerations, which were found to be lower in the bare frontal impacts than in their corresponding helmeted impacts (Figure 33c). This may be partially attributed to interference of the head surrogate nose with the impactor face in the bare frontal headform configuration.

Despite the reductions in peak linear acceleration and peak rotational rate in helmeted impacts at a given ram speed, it was observed that helmeted impacts produced higher primary strain peaks for similar peak linear accelerations and peak rotational rates (Figure 35a, and 35c). These trends were also observed in the secondary strain peaks (Figure 35b, and 35d), which were influenced by the neckform response, as the secondary peaks occurred after neckform engagement. The same trends were not observed when comparing peak MPS to peak rotational acceleration (Figure 35e, and 35f), which exhibited reductions in peak MPS for helmeted
impacts when compared to bare impacts at similar peak rotational accelerations for both the primary and secondary phases of headform motion.

![Graph showing time histories for the bare (solid) and helmeted (dashed) headform, shown for the 92nd percentile MPS in the (a) frontal and (b) rear impact orientations, and for the 95th percentile shear strain in the (c) frontal and (d) rear impact orientations.](image)

**Figure 34:** Time histories for the bare (solid) and helmeted (dashed) headform, shown for the 92nd percentile MPS in the (a) frontal and (b) rear impact orientations, and for the 95th percentile shear strain in the (c) frontal and (d) rear impact orientations.

The time-resolved strain data did not agree with any one of the particular kinematic measurements. Instead, the strain data in Figure 34 appeared to illustrate that helmets were more effective at reducing MPS at higher impact speeds than at lower impact speeds, and that helmets provided minimal reduction of shear
strain for the majority of impact speeds. For example, consider the 3.5 m/s and
the 5 m/s frontal impact configurations. For these impact speeds, peak rotational
accelerations were higher in the helmeted impacts (Figure 33) than in their bare
counterparts. However, the 92\textsuperscript{nd} percentile MPS values (Figure 34a) were lower in
the helmeted impacts than in comparable bare impacts. By contrast, in the 3.1 m/s
frontal impact configuration, the helmeted impact produced higher MPS than in
the corresponding bare impact, indicating a possible lower boundary to the impact
speeds for which this helmet is designed. Though the specific cause of these differ-
ing responses is not investigated further in this work, this comparison illustrates
the additional insight that can be gathered by evaluating and differentiating the
kinematic and MPS responses simultaneously using this methodology and head-
form.

The strain data extracted from this headform also illustrated the importance
of modelling head impacts beyond their initial kinematic response phase, as sec-
ondary phase MPS peaks were larger than the primary strain peaks in many of the
brain regions in the frontal impact configuration (Figure 27). This observation did
not apply to the rear impact configuration, which had consistently higher primary
MPS peaks (Figure 34b). The rear impact configuration also produced consistently
higher MPS than the frontal impact configuration at similar ram speeds, which is
consistent with observations made in literature [46,91,97].
Figure 35: Peak MPS plotted against peak kinematic measurements for the two phases of headform motion: peak linear acceleration for (a) primary, and (b) secondary phase of motion; peak rotational rate for (c) primary, and (d) secondary phase of motion; and peak rotational acceleration for (e) primary, and (f) secondary phase of motion.
5.2 Headform Validation and Calibration

The goal of the BIPED headform is to produce a calibrated strain response that is representative of the response of the human brain to impact. The biofidelity of the global head accelerations of the BIPED headform were previously investigated in a study by Li et al. [69], which found that the BIPED headform exhibited comparable acceleration magnitudes to the cadaveric impacts analyzed in their study. However, significantly longer acceleration pulse durations in the BIPED headform were observed when compared to cadaveric impacts [69].

To investigate the agreement between the BIPED headform and cadaveric brain tissue data, three bare frontal BIPED impacts were compared to three equivalent frontal PMHS impacts. An analysis of the displacement time histories was performed using CORAplus. CORAplus is a software that provides a rating of the agreement between two time-based signals, a reference signal and a simulation signal, using a combination of a corridor rating and a cross-correlation rating [98]. The corridor rating describes the agreement of the simulation signal using a range of data points that bound the reference signal [98]. The cross-correlation rating evaluates the agreement of the simulation signal to the reference signal with respect to its size, phase, and shape [98]. Results range between 0 and 1, with a rating of 1 representing a perfect correlation, and a rating meeting or exceeding 0.7 indicating good biofidelity [99].

The PMHS signals used in this analysis were created by calculating the average x- and y- displacements per frame for groupings of neighbouring markers residing in the same anatomical regions of the PMHS brain. In the BIPED headform, average x- and y- displacements were calculated from sub-divided regions of interest representing groupings of pixels within equivalent landmarked “brain regions”. Figure 36 illustrates the marker groupings and subdivided regions of
interest in the PMHS brain and the BIPED headform, respectively. In total, each impact speed (load case) analyzed 28 displacement signals, 14 x-displacement and 14 y-displacement. The analysis was performed for a 70 ms time interval. All CORA parameters were maintained at default settings. Table 7 summarizes the analyzed load cases and their resulting CORAplus scores.

<table>
<thead>
<tr>
<th>Load Case</th>
<th>Rating</th>
<th>Weight</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.0 m/s Frontal Impact</td>
<td>0.139</td>
<td>0.33</td>
</tr>
<tr>
<td>1.2 m/s Frontal Impact</td>
<td>0.124</td>
<td>0.33</td>
</tr>
<tr>
<td>2.1 m/s Frontal Impact</td>
<td>0.222</td>
<td>0.33</td>
</tr>
<tr>
<td><strong>Total Score</strong></td>
<td>0.162</td>
<td></td>
</tr>
</tbody>
</table>

Table 7: Results of the CORAplus analysis.

The results of this analysis (Table 7) indicate poor agreement between the x-
<table>
<thead>
<tr>
<th>Anatomical Brain Region</th>
<th>Average x-Displacement Rating</th>
<th>Average y-Displacement Rating</th>
</tr>
</thead>
<tbody>
<tr>
<td>Frontal</td>
<td>0.201</td>
<td>0.089</td>
</tr>
<tr>
<td>Parietal</td>
<td>0.137</td>
<td>0.201</td>
</tr>
<tr>
<td>Occipital</td>
<td>0.317</td>
<td>0.068</td>
</tr>
<tr>
<td>Insular</td>
<td>0.122</td>
<td>0.145</td>
</tr>
<tr>
<td>Cerebellum</td>
<td>0.193</td>
<td>0.234</td>
</tr>
</tbody>
</table>

Table 8: Results of the CORAplus analysis, broken down by brain region.

...
Figure 37: Comparison of marker displacements measured in the BIPED headform, reproduced from Figure 23 to marker displacements measured in a PMHS impact, adapted from Dutrisac et al. [87] and reproduced from Hardy et al. [81]. (a) Marker displacement field for a 3.1 m/s bare frontal impact in the BIPED headform, (b) marker displacements for a 3.0 m/s frontal impact in CO-103 [87], (c) marker displacement field for a 3.1 m/s bare rear impact in the BIPED headform, and (d) motion patterns for two NDT clusters for a bare, 3.1 m/s offset occipital in C408-T4 [81].

generated in a bare frontal impact (Figure 37) to displacement fields observed in a comparable PMHS study [87], we see general similarities in the rotational motion of the brain. However, the motion at the inferior surface of the brain surrogate is significantly greater than observations made by Dutrisac et al. [87]. In their work, it
was shown that tissue displacement in a PMHS brain subjected to impact is greater at the outer cortex (superior boundary) and significantly damped towards the inferior surface of the skull. This further suggests that the current BIPED skull base is not providing a sufficiently restrictive boundary condition. Comparisons were also made between the displacement field generated in the BIPED by a 3.1 m/s rear impact and neutral density target (NDT) cluster displacements in a 3.1 m/s occipital impact in a PMHS [81]. While the cadaveric impact experiments performed by Hardy et al. [81] were not conducted using a similar experimental setup, there were some general similarities observed in the displacements. For example, both displacement fields had elongated traces towards the posterior of the head, and had substantial vertical displacement at the outer cortex.

Recent investigations of displacement and strain-based head model validation techniques [64,65] have highlighted the importance of using strain data to validate models used for strain prediction, as models validated solely against displacement data do not necessarily produce accurate intracranial strain predictions [64, 65]. Given that PMHS strain data collected using the same impact configuration is not available at this time, a preliminary comparison will instead be made to DAMAGE model [61] strain predictions. While the DAMAGE model is subject to the limitations associated with the validation of the GHBMC FE model [61,63,65], it may serve as a preliminary comparison for the strain data measured in the BIPED headform. The time-resolved kinematic BIPED data was used to calculate the 95th percentile peak MPS using the DAMAGE model for the impacts described in Table 3. These are summarized alongside the BIPED-measured 92nd percentile primary and secondary peak MPS values in Table 9. It was found that the DAMAGE model reported consistently lower peak MPS estimates than the BIPED headform, with the highest agreement occurring for primary-phase MPS peaks in the bare
frontal impact configuration. The high strain in the BIPED brain surrogate is fur-
ther indicative of insufficient boundary conditions, and may suggest the need for
a stiffer brain surrogate material.

While the biofidelity of the current brain surrogate response to impact is lim-
ited, further refinement of the BIPED headform will ensure that future iterations
will provide a response that is adequately representative of cadaveric and clini-
cal datasets. Despite the current limitations, this technique and physical model
will serve as a valuable tool in the evaluation of headgear, allowing for a depar-
ture from rigid body kinematics in favor of experimentally generated intracranial
strain fields.
### Table 9: Comparison of peak MPS predicted by the DAMAGE model to peak MPS measured by the BIPED headform.

<table>
<thead>
<tr>
<th>Impact Config.</th>
<th>Impact Speed (m/s)</th>
<th>Bare Headform</th>
<th></th>
<th></th>
<th>Helmed Headform</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>DAMAGE Peak MPS</td>
<td>BIPED Peak MPS (0 - 35 ms)</td>
<td>BIPED Peak MPS (35 - 70 ms)</td>
<td>DAMAGE Peak MPS</td>
<td>BIPED Peak MPS (0 - 35 ms)</td>
<td>BIPED Peak MPS (35 - 70 ms)</td>
</tr>
<tr>
<td>Frontal</td>
<td>3.1</td>
<td>13.1%</td>
<td>15.0%</td>
<td>18.3%</td>
<td>11.5%</td>
<td>21.4%</td>
</tr>
<tr>
<td></td>
<td>3.5</td>
<td>15.3%</td>
<td>16.8%</td>
<td>22.0%</td>
<td>12.9%</td>
<td>22.0%</td>
</tr>
<tr>
<td></td>
<td>4.0</td>
<td>17.4%</td>
<td>20.8%</td>
<td>22.7%</td>
<td>14.6%</td>
<td>22.9%</td>
</tr>
<tr>
<td></td>
<td>4.5</td>
<td>19.5%</td>
<td>25.3%</td>
<td>36.7%</td>
<td>16.0%</td>
<td>24.3%</td>
</tr>
<tr>
<td></td>
<td>5.0</td>
<td>21.4%</td>
<td>28.1%</td>
<td>40.1%</td>
<td>16.7%</td>
<td>29.8%</td>
</tr>
<tr>
<td>Rear</td>
<td>2.4</td>
<td>10.8%</td>
<td>19.1%</td>
<td>16.6%</td>
<td>9.8%</td>
<td>18.2%</td>
</tr>
<tr>
<td></td>
<td>3.1</td>
<td>14.6%</td>
<td>27.2%</td>
<td>20.1%</td>
<td>12.9%</td>
<td>24.2%</td>
</tr>
<tr>
<td></td>
<td>3.4</td>
<td>16.5%</td>
<td>37.0%</td>
<td>27.5%</td>
<td>15.0%</td>
<td>28.7%</td>
</tr>
</tbody>
</table>
Chapter 6

Concluding Remarks

This document describes and evaluates an early iteration of a novel ATD that permits the direct resolution of intracranial strain during helmet testing. The measurement of tissue-level responses, when linked to clinical outcomes of injury, will have a tremendous potential to guide the development of improved head protection, reducing both the incidence and severity of concussion. The goal of the BIPED headform is to produce a calibrated and highly biofidelic response to impact based on comparisons to PMHS data conducted using the same impact configuration [87]. While this headform has not yet achieved a response that is perfectly representative of human brain tissue, the data and preliminary validation investigations presented in this work will help guide ongoing efforts to improve the biofidelity of the head and brain surrogate, playing a key role in the broader development of the BIPED headform.

This work has illustrated the importance of direct resolution of parameters of interest, as the simultaneous evaluation of multiple related metrics allowed for the observation of specific phenomena in the head surrogate. By directly comparing both the peak kinematic data and peak strain data throughout the testing series, this work illustrated that helmet-related reductions in the kinematic response did not explicitly result in reductions in MPS at all impact speeds. The helmet
was more effective at reducing MPS at higher impact speeds than at lower impact speeds. Though the cause of these differing responses has not been investigated in this work, the headform and methodology presented in this study will allow for an improved understanding of the interactions of the various components of head protection, enabling for a more effective helmet development cycle for the reduction of tissue strain.

Furthermore, by observing the time-resolved strain and kinematic responses, it was observed that in the frontal impact configuration, secondary strain peaks were consistently higher than the primary strain peaks. The same was not true for the rear impact configuration, suggesting the importance of capturing the full motion of the head surrogate, as the strain response maintains a longer active timescale than the kinematic response.

Observations of the complete shear strain distribution demonstrated a high-rate inversion of the shear strain, occurring at rates significantly higher than the initial loading phase, in both bare and helmeted impacts. This inversion coincided with neckform engagement, which would have influenced the specific rates and peaks of the inversion. However, as the human neck would eventually restrict head motion following an impact, this phenomenon should be investigated further.

Additionally, the direct comparison of the displacement fields in the BIPED head surrogate allowed for a more clear understanding of the differences in the strain fields based on impact configuration, providing a clear visualization of the effects of impact location on the resulting brain surrogate motion.

**Recommendations and Future Work**

The use of an XDIC methodology allowed for a closed-head model and the addition of a fluid boundary condition (perfusion), eliminating issues seen in other
strain-measuring head surrogate models \cite{71} such as decoupling of the brain surrogate from the skull surrogate, and surface Rayleigh waves. However, analysis of the biofidelity of the displacement data suggested that the current boundary conditions in the BIPED headform are insufficient.

Though changes to boundary conditions are a complex task, the reincorporation of a tentorium surrogate would be advisable given its inclusion in other BIPED iterations. While it may introduce a strain concentration \cite{71} in the surrounding tissue, it would also serve to restrict motion at the inferior surface of the skull. The intention would be to damp motion in the inferior portion of the brain surrogate, allowing for a displacement field more in line with observations made by Dutrisac et al. \cite{87}.

Other anatomical simulating structures may be explored to better mimic the boundary conditions present in the human head, such as a brain stem or meninx layer, as these structures provide resistance to the motion of the human brain.

Furthermore, as both the displacement and strains within the brain surrogate appeared to be excessively high, an investigation of brain surrogate material choice would be valuable. An increase in material stiffness may serve to reduce the brain surrogate motion sufficiently with minimal changes to the existing boundary conditions.

Finally, given the implications of the neckform on the key observations made in the analysis of the shear strain in the headform, a thorough investigation on the biofidelity of the unbiased neckform and the effects of neckforms on intracranial strain fields is suggested.

While the head surrogate has not yet achieved a response that is perfectly biofidelic, further calibration of the headform will ensure that it is adequately representative of PMHS and clinical data. Once calibrated, this technique will serve as a valuable tool in the evaluation of headgear through experimental intracranial
strain fields, allowing for improved head protection that will reduce the incidence and severity of concussion.
List of References


[31] E. Becker, “Helmet development and standards,” Frontiers In Head And Neck Trauma Clinical and Biomedical, 1998.


