Investigation of Intracranial Brain Deformation in Post-Mortem Human Subject Heads Due to Blunt-Force Impacts

J Demiannay, M Brannen, S Xu, A H Azadi, P Dehghan, S Goodwin, W Yahiaoui-Djerboua, R Taylor, K Poon, S Brien, M Vassilyadi, M D Cusimano, C Karton, T. B Hoshizaki, O E Petel

I. INTRODUCTION

Traumatic brain injury (TBI) is prevalent in sports, workplace, motor vehicle, and military incidents, and can result in both short- and long-term health effects [1]. Mild traumatic brain injury (mTBI) is increasingly attributed to rotational motion and accelerations causing brain tissue deformation, resulting in diffuse injuries across the parenchyma of the brain [2–3]. Finite element (FE) models are a powerful tool for developing mTBI injury risk criteria [4], however, real-world biofidelic brain tissue deformation datasets required for model validation remain limited [5–8]. This experimental study investigated brain biomechanics in post-mortem human subject heads, measured brain-skull relative deformation, examined the effects of impact conditions on the dynamic response of the brain, and developed biofidelic datasets that can be implemented in FE model validation.

II. METHODS

Three specimens were subjected to a series of 5 to 12 sagittal impacts: CO-103 and CO-109 were subjected to frontal and occipital impacts respectively, using a less compliant modular elastomer pad (MEP) impactor (60 shore A), while CO-108 was subjected to frontal impacts using a more compliant 2-layer vinyl nitrile (VN) foam impactor (53 Shore A). Skull-brain relative displacements and strains were measured by recording the motion of embedded radiopaque markers using a high-speed X-ray imaging (HSXR) system.

Each specimen was fitted with a mechanical neck fixture without directional bias, enabling impacts in an upright configuration. Pre-test MRIs were used to ensure no brain abnormalities were present that would affect the mechanical response of the brain, and to pre-plan marker placement. Carleton Elastomeric Markers (CEM) were designed for optimal stiffness and density to minimise the mechanical influence on the surrounding brain tissue [8]. Two marker planes were used; a primary marker plane was located 20 mm parasagittal, and a secondary marker plane was located 5 mm parasagittal and offset from the primary plane by 7.5 mm, targeting the corpus callosum and brain stem. Post-impact MRIs were used to confirm marker placement, and to identify the anatomic localisation of each marker. The carotid and vertebral arteries, ventricles, and subdural space of each specimen were perfused with artificial cerebrospinal fluid (aCSF) to improve biofidelity.

Blunt-force impacts were delivered using a 13 kg pneumatically driven linear impactor (Cadex Inc.) at nominal impact velocities between 2 to 4 m/s. Impacts were captured using a HSXR imaging system at 5000 fps. Head kinematics were recorded at 20 kHz using a 6DX Pro Slice kinematic sensor package (DTS Inc.) rigidly attached to the neck and filtered using a CFC 180 (300 Hz) low-pass filter. Data processing and marker tracking were performed using custom MATLAB programs to compute skull-brain relative displacement and strain in a skull-fixed reference frame. Embedded markers served as nodes for 2D triangular strain elements, and Green-Lagrange strains were calculated using markers embedded within the same anatomic regions.

III. INITIAL FINDINGS

Table 1 summarises the resulting peak linear and rotational accelerations for each specimen at a nominal impact velocity of 3 m/s. Impact conditions, primary plane marker locations, and resultant displacement profiles

J. Demiannay, M. Brannen, and S. Xu are PhD candidates, and O. E. Petel is an Associate Professor in Mechanical and Aerospace Engineering at Carleton University in Canada (jeanjacquesdemiannay@cmail.carleton.ca; +1-613-261-3124). A. H. Azadi, P. Dehghan, and C. Karton are Research Associates, and T. B. Hoshizaki is a Professor of Human Kinetics in the Department of Health Sciences at the University of Ottawa in Canada. S Goodwin is a Senior Prosector, and W. Yahiaoui-Djerboua is a Manager at the University of Ottawa Anatomy Teaching Laboratory in Canada. R. Taylor is a Physicist at the Royal Ottawa Mental Health Center in Canada. K. Poon is a Neurosurgeon at the Clinique Neuro-Outaouais in Canada. S. Brien is a Senior Vice President Medical, Academic, and Research Affairs at Horizon Health Network in Canada. M. Vassilyadi is a Neurosurgeon in the Department of Surgery at The Ottawa Hospital in Canada. M. D. Cusimano is a Neurosurgeon in the Department of Surgery at St. Michael's Hospital in Canada.

for six markers in each specimen in comparable locations are displayed in Fig. 1. Specimens CO-103 and CO-109, both subjected to less compliant impacts, exhibited linear and rotational accelerations more than double those of CO-108 for similar impact velocities (Table 1). In all specimens, the largest displacements were measured in the superior regions of the brain, namely the frontal and parietal lobes, compared to more inferior regions. Occipital impacts consistently resulted in the lowest skull-brain relative displacements compared to frontal impacts of similar impact velocities.

TABLE 1 IMPACT CONDITIONS AND RESULATANT LINEAR AND ANGULAR ACCERATION FOR CO-103, CO-108, AND CO-109.

CO-103 (MEP, 60 Shore A)			**CO-108 (2-Layer VN foam, 53 Shore A)**			**CO-109 (MEP, 60 Shore A)**		
Impact Velocity (m/s)	Acceleration Linear (g)	Acceleration Angular (rad/s^2)	Impact Velocity (m/s)	Acceleration Linear (g)	Acceleration Angular (rad/s^2)	Impact Velocity (m/s)	Acceleration Linear (g)	Acceleration Angular (rad/s^2)
3.0	62	3754	3.0	39	1551	2.9	109	3972

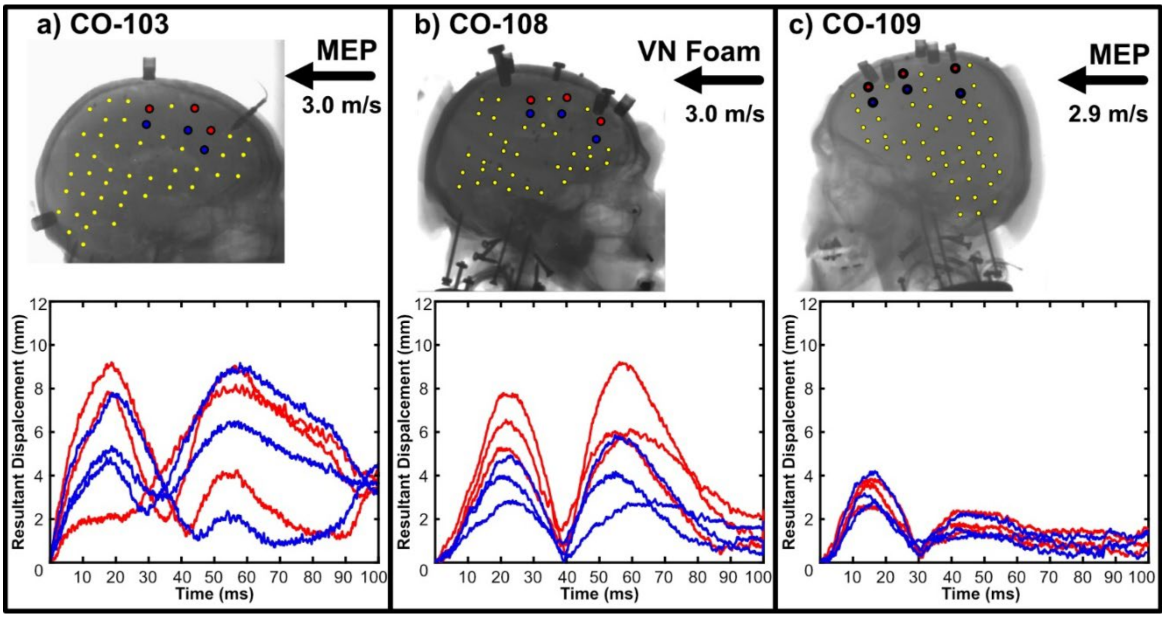

Figure 1 Impact conditions and resultant displacement profiles for specimens CO-103, CO-108, and CO-109 for a 3 m/s impact. Highlighted markers show primary plane marker locations. Six makers in similar locations were selected in each (highlighted in red and blue), with the resultant displacement profiles of each marker being displayed.

IV. DISCUSSION

Peak skull-brain relative displacement and strain increased with impact velocity across all PMHS specimens, with the largest displacements and strains occurring in the frontal lobe, irrespective of impact direction. Comparison of impact direction, compliance, and kinematic response underscores the complexities of the dynamic response of the brain to blunt-force trauma. Notably, CO-109 exhibited the lowest intracranial displacements and strains despite generating the largest linear and rotational accelerations compared to frontal impacts of similar impact velocities, regardless of impact compliance. One might expect that larger accelerations would lead to larger deformations; however, the unique geometry of the skull, complex intracranial anatomic features, and mechanical and physiological properties of brain tissue result in a dynamic response that is difficult to predict. This study provides real-world biofidelic skull-brain relative displacement and strain datasets that can be used to advance understanding of brain biomechanics, and support FE head model validation.

V. REFERENCES

[1] Nguyen R *et al.*, Can J Neurol Sci, 2016.
[2] Knutsen A *et al.*, Brain Multiphys, 2020.
[3] Zhan X *et al.* , Biol Med, 2025.
[4] Ji S *et al.*, Ann Biomed Eng, 2014.
[5] Hardy W *et al.*, Stapp Car Crash J., 2007.
[6] Guettler A *et al.*, SAE Inernational, 2018.
[7] Alshareef A., J Neurotrauma., 2020.
[8] Dutrisac S., Ann Biomed Eng, 2021.