# Application of Measured Head Kinematics to Finite Element Head Model and Assessment of Brain Response in Human Volunteers Subject to Acceleration Events from Rifle Recoil

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## I. INTRODUCTION

Mild traumatic brain injury (mTBI) has been recently proposed as a potential occupational risk among Canadian Armed Forces (CAF) service members due to repeated exposure to rifle recoil events [1]. When firing, the recoil forces are transferred through the shoulder and neck to the head, causing head accelerations. As such, it is important to understand the head and brain response to such accelerations and the possible risk of injury. The measurement of gross head kinematics and, more recently, brain tissue deformation using detailed finite element (FE) head models are commonly used to assess the risk of brain injury resulting from various types of loading condition. Maximum principal strain (MPS), a common brain deformation metric, has been identified as an indicator of injury risk , with a higher correlation to concussion than head kinematics [2]. Although MPS has been widely used to assess brain response [3], this metric is limited as a single value and is typically extracted from the entire brain. Thus, the response of brain regions with different strain thresholds associated with concussion may be lost [4]. In the present study, a FE head model was used to assess the sensitivity of two brain response metrics, maximum principal strain (MPS) and a newly proposed cumulative strain measure [5], to detect differences in brain deformation between three acceleration events resulting from the firing of three different long-range rifles. Head kinematics were experimentally measured from a single volunteer subject during firing, and used as input conditions to the FE head model to assess the resulting brain response (Fig. 1).

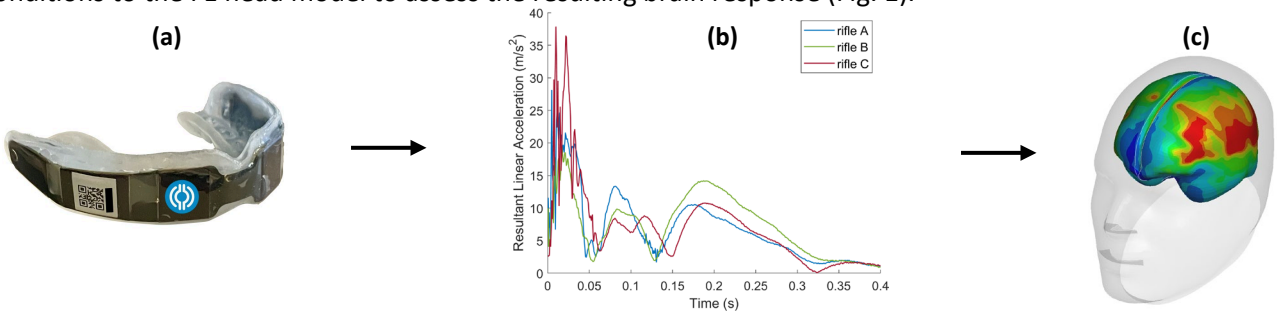


Fig. 1. An overview of the analysis process starting with (a) data collection using instrumented mouthguards. (b) Measured head kinematics including resultant linear acceleration, one per rifle, used as inputs to the (c) GHBMC 50$^{th}$ adult male head model to assess brain deformation and injury risk.

## II. METHODS

The volunteer subject (UW REB #44306) fired three different rifles (labelled A, B and C in this study) four times. Angular velocity and linear accelerations were recorded using instrumented mouthguards (Prevent Biometrics, USA) and processed in MATLAB. Raw data from the mouthguards were transformed into the J211 coordinate system. Fourth-order low-pass filters with 500 Hz and 50 Hz corner frequencies were applied to the linear acceleration and angular velocity data, respectively. A rigid body kinematic transform (Eq. 1) was used to compute linear acceleration at the centre of gravity (CoG) of the head, to be used as an input for the computational head model. Angular velocity, the other input to the head model, did not require a kinematic transform.

$$\boldsymbol{a_{head}} = \boldsymbol{a_{mouthguard}} + \boldsymbol{\alpha} \times \boldsymbol{r} + \boldsymbol{\omega} \times (\boldsymbol{\omega} \times \boldsymbol{r}) \quad (1)$$

where $a_{head}$ is linear acceleration at head CoG, $a_{mouthguard}$ is linear acceleration recorded by mouthguard, $\alpha$ is angular acceleration, $\omega$ is angular velocity, and $r$ is the distance from mouthguard to head CoG scaled using anthropometric data of the volunteer subject. Angular acceleration ($\alpha$) was calculated by taking the derivative of experimental angular velocity. The GHBMC 50$^{th}$ percentile male head model [2] was used to simulate each experimental test, for a total of 12 simulations. Linear accelerations and angular velocities associated with each test were inputted to the CoG of the head model using prescribed rigid body motions assigned to the skull following previous studies [4]. The models were run on a high-performance computing cluster and solved using a

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commercial FE code (LS-DYNA R9.2 single-precision, ANSYS, Canonsburg). Brain tissue strains were extracted from the model for eight brain regions that had previously been associated with concussion. MPS was extracted from the simulations for each brain region. Cumulative strain plots were constructed by calculating the MPS of each element throughout the simulation time as well as the initial volume fraction of each element. Next, the elements were ordered in ascending order of MPS and the fraction of volume of each element was subtracted from the initial volume of 100%. Cumulative strain curves were then generated by plotting MPS on the x-axis and percentage volume on the y-axis to show the distribution of MPS within a brain region with respect to the volume of that region. The area under the cumulative strain plot ($A_{CS}$) was integrated with respect to percentage volume (y-axis) for statistical analysis.

## III. INITIAL FINDINGS

The measured kinematics exhibited visual differences between each rifle. The mean MPS value was calculated from each simulation and brain region (Fig. 2a, showing results for the cerebellum). The mean MPS values for the cerebellum were 0.0313, 0.0306 and 0.0364 for rifles A, B and C, respectively (Fig. 2a). The cumulative strain distribution of the cerebellum for each shot taken by the volunteer subject was plotted (Fig. 2b). The mean area under the cumulative strain plots for rifles A, B and C were 0.0109, 0.0098 and 0.013, respectively. A one-way ANOVA with a significance level of 0.05 was conducted to assess differences between the three rifles using MPS values and areas under the cumulative strain plots. The mean MPS values were not statistically different among the rifles, with a P-value of 0.171 (> 0.05). However, the mean areas from the cumulative strain plots were statistically different, with a P-value of 0.000579 (< 0.05).

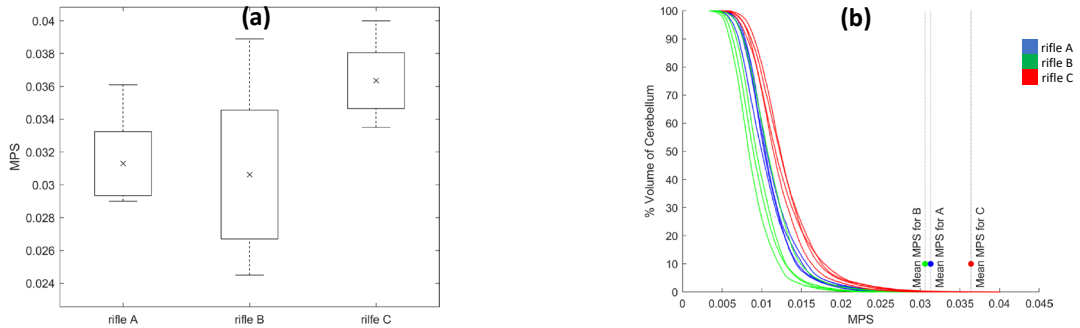


Fig. 2. a) Mean MPS and variation for the three rifles and b) cumulative strain plots for each shot across the three rifles. The circular markers indicate the location of the mean MPS values.

## IV. DISCUSSION

In this study, MPS was less sensitive than cumulative strain plots and $A_{CS}$ in detecting differences in tissue deformation between the three acceleration cases, as MPS condenses a large amount of data for a brain region to a single value. Additionally, as observed in the cumulative strain plot (Fig. 2b), much less than 5% of the cerebellum volume experienced high levels of strains (>0.0313), indicating that the single-value MPS metric did not represent the overall deformation over the volume of the cerebellum. In contrast, the cumulative strain plots provided a detailed representation of the strain distribution in each brain region, leading to an improved understanding of variations within a particular acceleration event and differences between acceleration events. Furthermore, the cumulative strain plots (Fig. 2b) showed a clear trend between the three acceleration events, both visually as well as statistically. Previous studies have reported injurious strain thresholds ranging from 0.1 to 0.47 for MPS [6]. The highest MPS observed in this case was 0.04, which is below reported thresholds. While no acute injuries occur from a single event, the accumulation of these events over time can lead to symptoms. The collection of head kinematics data from a controlled population to be used an input to a FE model is a promising methodology. A study on a larger population is ongoing and will provide important data to evaluate brain response metrics further.

## V. REFERENCES

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