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Instantaneous Whole-brain Pressure Estimation in Translational Head Impact via Pre-computation

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Abstract:

Background**:** Finite element models are effective tools to study brain injury in automotive crashes due to their unique capability in providing tissue-level responses that is otherwise infeasible in a crash dummy. However, achieving an optimal tradeoff between model sophistication and computational efficiency is a practical challenge.

Objective**:** In this study, we intend to propose a pre-computation scheme and present a simple, yet effective technique to instantly estimate whole-brain pressures induced by linear acceleration.

Methods**:** Three hydrostatic atlas pressure responses along the three major axes of the head were pre-computed. Whole-brain pressures were then superimposed in real-time for an arbitrary translational head impact. The accuracy of the pressure estimation scheme was tested in seven real-world impact cases ranging from cadaveric experiments to reconstructed automotive crashes.

Results**:** This study successfully employed a simple and effective pre-computation strategy to estimate whole-brain pressures in real-time for translational head impacts (<0.1 sec on a lap-top vs. ~55 minutes for the overall processing on a high-end workstation). The technique was efficient and sufficiently accurate for most cases as long as the impulse duration is greater than 2 ms, demonstrating its high effectiveness for real-world applications.

Conclusions**:** This technique may have the potential to aid kinematics-based injury metrics such as head injury criterion (HIC) to more effectively assess the risk and severity of pressure-induced coup/countercoup or moderate to severe brain injuries in the future.

Keywords: Traumatic brain injury, translational head impact, linear acceleration, pre-computation, Dartmouth Head Injury Mode

1 Introduction

Traumatic brain injury (TBI) continues to be a significant societal problem in the face of rapid increase in motorization without significant improvement in road safety. Understanding the biomechanical mechanisms of TBI is critical to establish effective injury diagnostic criteria as well as to develop protective gears and rules to prevent and reduce the incidence and severity of the injury.

Historically, kinematics-based injury criteria, such as a_{3ms} (maximum 3 ms-average of the resulting linear acceleration (a_{lin}) to be less than a given threshold), Head Injury Criterion (HIC) have been extensively applied in automotive industry to assess the potential injury risk. Unfortunately, these global kinematic variables and their variants do not directly relate actual brain injuries to tissue-level mechanical responses believed to initiate the injury [1–3]. Therefore, their effectiveness in assessing the risk of TBI has been criticized.

Early efforts to understand the biomechanical basis of TBI have focused on the correlation between tissue-level pressure responses and the measured a_{lin}. It was found that pressure throughout the brain surrogates during an impact event showed a good correlation between peak pressure and peak magnitude of a_{lin} (a_{lin}^{p}) [4,5]. For the living brain, several studies found that the transient pressure increase within the brain could cause neurological dysfunction [6–9]. However, brain pressure responses cannot be exported from crash dummy used in vehicle safety design. As a result, finite element models have become increasingly important to bridge the gap between macro- and micro-scale TBI biomechanics studies [2,10].

Developing a computationally efficient, yet sophisticated numerical model with high predictive power remains a major challenge [11]. To address this, we have proposed a pre-computation strategy for (near) real-time brain strain estimation in the context of sports-concussion. Essentially, this technique establishes a large "look-up" table based on characteristics of measured on-field rotational accelerations, a_{rot} [12]. Instead of directly simulating each head impact, the pre-computed brain response atlas (pcBRA) interpolates or extrapolates brain strain responses within seconds or instantly without significant loss of accuracy from the estimation itself. Here in this study, we extend the pre-computation strategy to rapidly estimate whole-brain pressure responses from a_{lin}. Seven typical head impact scenarios were chosen from cadaveric experiments and reconstructed car crashes to evaluate the accuracy performance of the technique.

2 Methods

2.1 The Dartmouth Head Injury Model

All head impact simulations were conducted using the Dartmouth Head Injury Model (DHIM; Figure.1). The details of model creation, material property assignment, and successful validations have been reported recently [13–15]. The DHIM is composed of solid hexahedral and shell quadrilateral elements with a total of 101.4 k nodes and 115.2 k elements (element size of 3.2±0.94 mm) and a combined mass of 4.562 kg for the whole head (1.579 kg or 1.558 kg for the brain with or without the spinal cord). The DHIM has been successfully validated against relative brain-skull displacement [16,17] and intracranial pressure responses [18,19] from cadaveric experiments, as well as full-field strain responses in a live human volunteer [20], with an overall "good" to "excellent" validation performance.

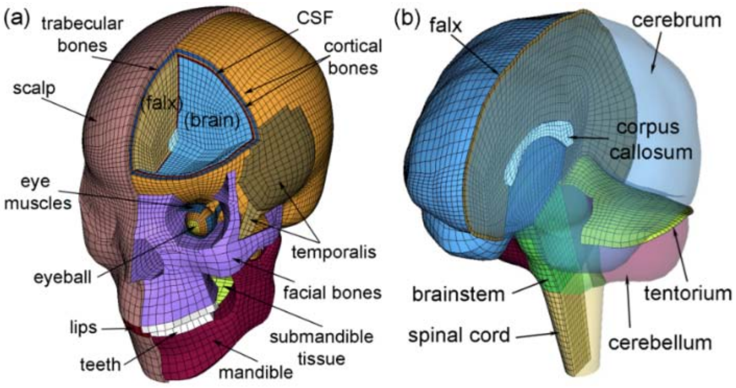

Figure.1 The DHIM showing color-coded head exterior (a) and intracranial components (b), which also includes part of the spinal cord to improve model biofidelity in the inferior region. The *x*-, *y*- and *z*-axis of the model coordinate system corresponds to the posterior-anterior, right-left, and inferior-superior direction, respectively.

2.2 Atlas pressure responses

According to a recent dimensional analysis [15], brain pressure responses in a translational head impact are essentially a result of contact forces generated from relative normal brain-skull displacements at the interface, which are uniquely determined by a_{lin} (magnitude and directionality), brain size and shape. For a constant acceleration field of a_{lin} with an arbitrary direction, the resulting whole-brain pressure, p, is hydrostatic. The corresponding brain-skull displacement, d, is also a constant, which can be decomposed into three orthogonal components, d_x, d_y, and d_z, respectively, along the three major axes of the head model coordinate system:

$$\mathrm{d} = \mathrm{d}_x\mathrm{i} + \mathrm{d}_y\mathrm{j} + \mathrm{d}_z\mathrm{k}. \tag{1}$$

Because pressure is a scalar value, brain pressures corresponding to the three independent displacement components, p_x, p_y, and p_z, respectively, can be simply superimposed. This establishes the brain pressure responses corresponding to d or effectively, the given a_{lin}.

$$p = p_x + p_y + p_z. \tag{2}$$

As pressure is linearly related to the magnitude of a_{lin} and d, a set of baseline pressures corresponding to impacts along the three major axes, p'_x, p'_y, and p'_z, respectively, can be pre-computed. They can then be individually scaled linearly to obtain pressures corresponding to d_x, d_y, and d_z, or effectively, p_x, p_y, and p_z, respectively. Here, we chose a baseline peak a_{lin} magnitude of 100 g (albeit, the magnitude is not important; 1 g = 9.8 m/s2) to establish the three baseline pressures using the DHIM. This leads to the following set of equations: /100

$$p_y = a_y \times p'_y/100, \text{ and} \tag{3}$$

$$p_z = a_z \times p'_z/100$$

where a_x, a_y and a_z are the a_{lin} magnitude components along the three major axes (in g) at a given time. Combining Eqns. 2 and 3, whole-brain pressure can be obtained using the following equation:

$$p = (a_x \times p'_x + a_y \times p'_y + a_z \times p'_z)/100. \tag{4}$$

It is important to note that the above analysis considers a constant acceleration. For any real-world time-varying a_{lin} temporal profile, the resulting hydrostatic pressure at an arbitrary time instance essentially generates a first-order approximation, purposefully neglecting any potential influence from the dynamic inertial effect. Nevertheless, the hydrostatic response (Eqn. 4) would be sufficiently accurate for most real-world impacts because brain pressure reaches an equilibrium almost instantly due to its high dilatational speed (e.g., with impulse duration >2 ms according to [21]).

To establish the atlas hydrostatic pressures (i.e., p'_x, p'_y, and p'_z in Eqns. 3 and 4), a "ramp-and-hold" a_{lin} profile (Figure.2) was applied to the rigid skull separately along the three major axes. Brain pressures averaged within the 15–20 ms window were used to serve as the atlas responses. This particular temporal profile was used to ensure reaching a hydrostatic response and to avoid perturbations from the dynamic inertial effects (arrow in Figure. 2). The peak magnitude of a_{lin} (i.e., 100 g) was not important as the pressure responses would be linearly scaled. Importantly, the resulting element-wise, hydrostatic pressures served as atlas "modes" for linear scaling and superposition (Eqn. 4).

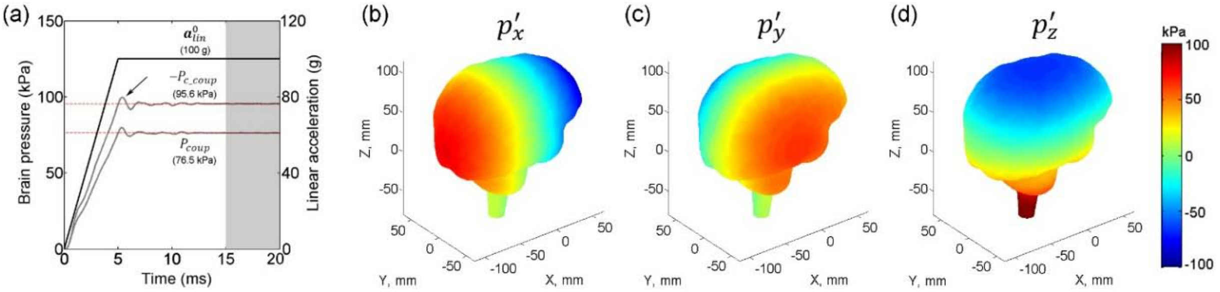

Figure.2 a_{lin} profile and the resulting P_{coup} and P_{c_coup} for an occipital impact (i.e., along the *x*-axis) (a) as well as the three alas pressure responses (b–d). Element-wise pressure values within the 15–20 ms window (shaded area; (a)) were averaged to serve as the atlas response. Pressure perturbations due to dynamic inertial effects are evident (arrow).

2.3 Evaluation of estimation accuracy

Seven 3DOF a_{lin} profiles ranging from cadaveric head impacts to reconstructed traffic accidents were used for evaluation. Their a_{lin} temporal profiles were manually digitized from their respective published sources. The a_{lin} magnitudes ranged 52.3–680.2 g and the HIC ranged 225–6606.7 (Table 1).

Table 1. Summary of impact scenarios, injury details, HIC and the a_{lin} peak magnitudes (a^p_{lin}) for the cases evaluated.

Case #	Impact scenario	Injury details	HIC	a^p_{lin} (g)	Reference
1	cadaveric experiments	no injury	786.2	199.2	[18]
2		N/A (not reported)	226.2	91.0	[19]
3	reconstructed traffic accidents	AIS 4: GCS = 11 (moderate)*	6606.7	680.2	[22]
4		AIS 5: extradural haematoma; contusion, GCS = 3 (severe)*	1782.2	217.6	
5		AIS Multiple: extradural and subdural haematoma; cerebral swelling and contusion	2221.1	293.5	
6		N/A (not reported)	668	90.0	[23]
7			225	52.3	

AIS: Abbreviated Injury Scale; GCS: Glasgow Coma Scale; *: According to [24]

2.4 Data analysis

All brain pressure response simulations were obtained from the DHIM via Abaqus/Explicit (Version 6.12; Dassault Systèmes, France). The typical runtime for a 40 ms impact was ~50 minutes plus ~5 min for post-processing on a high-end multi-core Linux cluster using 8 CPUs (Intel Xeon X5560, 2.80 GHz, 126 GB memory). With the pcBRA, whole-brain pressure estimation for an entire a_{lin} profile took <0.1 sec on a laptop (essentially the estimation process is a simple matrix multiplication). All data analyses were performed in MATLAB (R2015a; MathWorks, Natick, MA).

3 Results

Figure.3 illustrates the pressure estimation process for one case. Figure.4 graphically compares pressure distributions between the estimated and the directly simulated responses for two additional cases. For the remaining cases, the estimated P_{coup} and P_{c_coup} temporal profiles are compared with their directly simulated counterparts, along with the resultant a_{lin} profiles (Figure.5). For six out of seven cases, the differences between the estimated and directly simulated peak pressures were within 10% of the directly simulated peak responses, and were considered successful. Their percentage differences in P_{coup} and P_{c_coup} ranged from 0.0–9.2% (5.4±3.9%) and 0.3–9.0% (5.1±3.8%), respectively. One case failed, which had a percentage difference in P_{coup} and P_{c_coup} of 27.8% and 23.4%, respectively.

4 Discussion

This study successfully employed a simple and effective pre-computation strategy to estimate whole-brain pressures in real-time for translational head impacts (<0.1 sec on a lap-top vs. ~55 minutes for the overall processing on a high-end workstation). Using the pressure superposition law, the three pre-computed hydrostatic pressures along the three major axes of the DHIM were directly used to estimate whole-brain pressure for an arbitrary a_{lin} profile at any instance in time. Neglecting the dynamic inertial effect essentially led to a first-order approximation of the estimated responses.

Regardless, even the first-order approximation was sufficiently accurate for six out of seven real-world 3DOF a_{lin} profiles. Peak P_{coup} and P_{c_coup} differed from their directly simulated counterparts by 5.4% and 5.1% on average, respectively, for all the successful cases pooled. These relative errors were far below those in real-world a_{lin} measurement (e.g., up to 17–31% according to [25,26]).

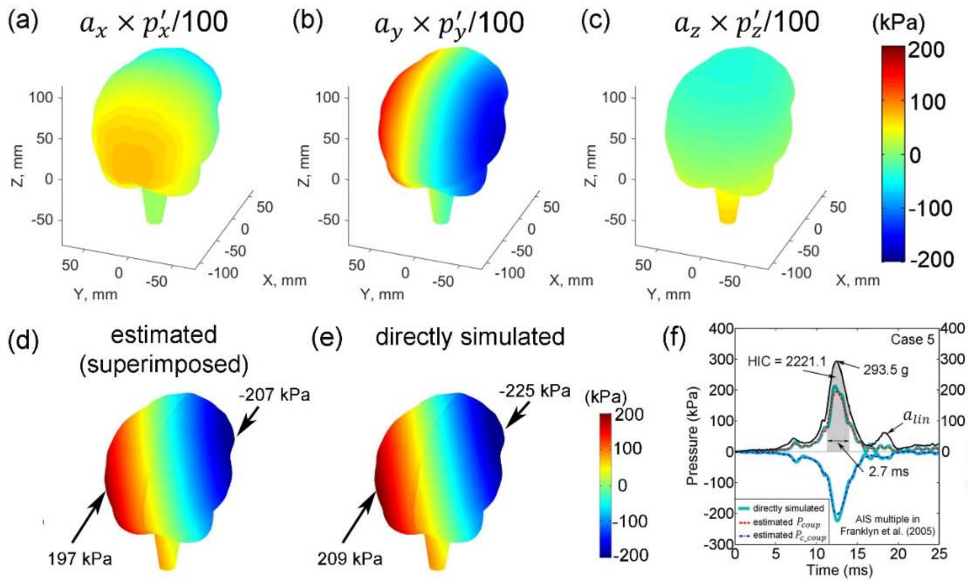

Figure.3 Illustration of using three atlas responses to estimate a pressure distribution map for an arbitrary given time in a selected a_{lin} profile (case 5). The atlas responses are first properly scaled (a–c) and then superimposed (d), resulting in a nearly identical distribution relative to the directly simulated counterpart (e). The estimated P_{coup} and P_{c_coup} overlaid with resultant a_{lin} profile were shown in (f; time interval corresponding to HIC shaded).

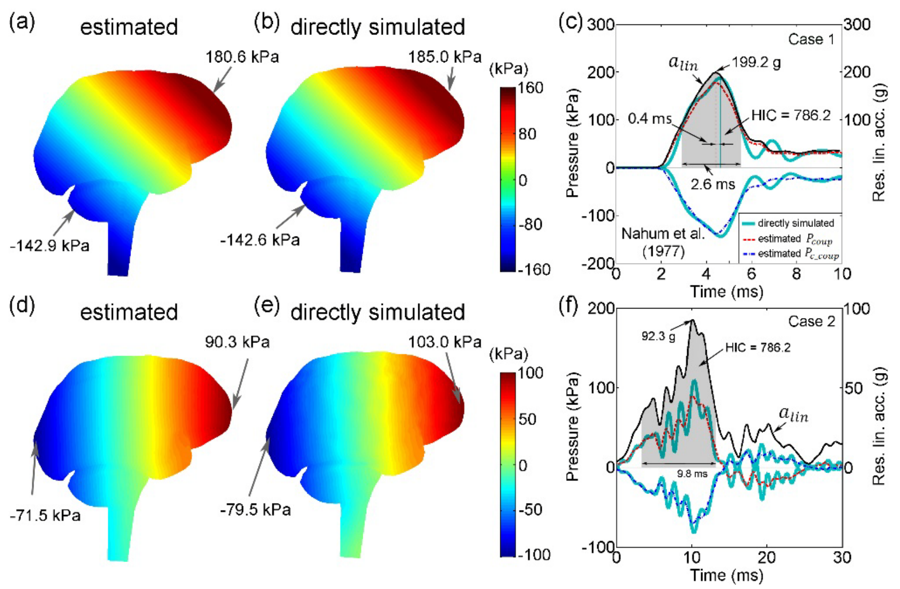

Figure.4 Comparison between the estimated (left) and directly simulated (middle column) whole-brain pressure distributions when P_{coup} reached its peak for two additional case. The response temporal profiles (estimated and directly simulated are nearly identical) are compared with that of the resultant a_{lin} (right).

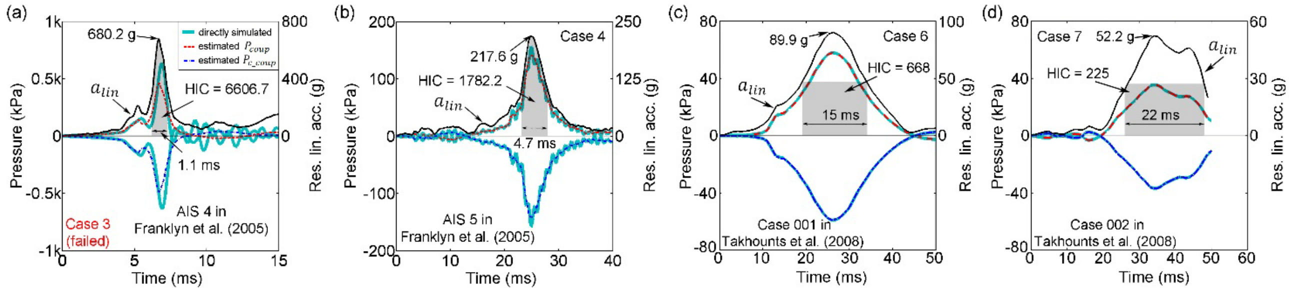

Figure.5 Comparison between the estimated and directly simulated pressure temporal profiles for the remaining four cases, along with the resultant a_{lin} profiles.

The failed case (#3) had an extremely short impulse (~1 ms) than others (≥3-4 ms; Figure. 6c), although it also corresponded to the largest a_{lin} and HIC magnitudes. The estimated peak P_{coup} and P_{c_coup} were significantly lower than the directly simulated counterparts (by 27.8% and 23.4%, respectively; albeit, still below the maximum a_{lin} measurement error; Figure.3), which was a result of the dynamic inertial effects not captured by the hydrostatic atlas pressures. Such an pressure magnification by dynamic inertial effect was also observed in [27], in which an identical kinetic energy was delivered to a head model to generate a range of impact durations. Therefore, caution must be exercised when estimating pressures for extremely short-duration impact.

Most real-world head impacts last longer than 2 ms [21]. Therefore, the pressure estimation scheme would be sufficiently accurate. However, very short a_{lin} impulses (e.g., < 5 ms) are more likely to be related to focal injuries such as subdural hematoma, epidural hematoma, subarachnoid hemorrhage and contusion [28]. Consequently, there is a need to improve the pcBRA performance further in these situations. Again, a pre-computation strategy may be effective by densely sampling the causal relationship between impact and the resulting pressures to allow pressure compensation. This is conceptually analogous to our pre-computation strategy for brain strain responses induced by a_{rot} in contact sports [12,29].

Another potential limitation with the study was that the atlas pressure responses were derived from a rigid-skull assumption, which allows the estimated pressures to serve as an upper bound when non-rigid skull deformation becomes significant [15]. On the other hand, when significant non-rigid skull deformation occurs in an extremely short duration (e.g., Case 3 in Figure. 5), competition of pressure over- and under-estimation may ensue. The overall effect on pressure responses needs further investigation.

In addition, head rotations always occur in real-world impact, but this was not considered in this study. To compensate for rotation-induced pressure, we have developed a pressure superposition to incorporate pressures induced by both a_{lin} and a_{rot}. Initial results suggest favorable performance when using full degree-of-freedom, real-world head kinematics [30].

Despite these limitations, this study successfully established a simple, yet effective method to estimate brain pressures in real-time in realistic translational head impacts. Future work will investigate whether this technique has the potential to aid other kinematics-based injury metrics such as HIC to improve the assessment of head injury, thereby to improve vehicle safety designs in the future.

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References

[1] Morrison III, B., Cullen, D., and LaPlaca, M., In vitro models for biomechanical studies of neural tissues. Neural Tissue Biomech. Stud. Mechanobiol. Tissue Eng. Biomater.. 2011. 3: p. 247–285.
[2] Meaney, D. F., Morrison, B., and Bass, C. R, The Mechanics of Traumatic Brain Injury: A Review of What We Know and What We Need to Know for Reducing Its Societal Burden. J. Biomech. Eng., 2014. 136(2): p. No. 021008.
[3] King, A. I., Yang, K. H., Zhang, L., Hardy, W., and Viano, D. C., Is head injury caused by linear or angular acceleration? in IRCOBI Conference. 2003. Lisbon, Portugal
[4] Gurdjian, E., Lissner, H., Evans, F., Patrick, L., and Hardy, W., Intracranial pressure and acceleration accompanying head impacts in human cadavers. Surg. Gynecol. Obstet.,1961. 113: p. 185–190.
[5] Thomas, L. M., Roberts, V. L., and Gurdjian, E. S., Experimental intracranial pressure gradients in the human skull. J. Neurol. Neurosurg. Psychiatry., 1966, 29(5): p. 404–11.
[6] Gurdjian, E., Lissner, H., Webster, J., Latimer, F., and Haddad, B., Studies on experimental concussion: relation of physiologic effect to time duration of intracranial pressure increase at impact. Neurology, 1954. 4(9): p. 674–681.
[7] Lindgren, S., and Rinder, L., Experimental studies in head injury. II. Pressure propagation in 'percussion concussion'. Biophysik, 1966. 3(2): p. 174–180.
[8] Denny-brown, D., and Russell, W. R., Experimental cerebral concussion. Brain,1941. 64(2-3): pp. 93–164.
[9] Gurdjian, E., Webster, J., Lissner, B., Haddad, F., and Latimer, H., Quantitative determination of acceleration and intracranial pressure in experimental head injury; preliminary report. Neurology, 1953. 3(6): pp. 417–423.
[10] Yang, K. H., Mao, H., Wagner, C., Zhu, F., Chou, C. C., and King, A. I., Modeling of the Brain for Injury Prevention. Neural

Tissue Biomechanics: Studies in Mechanobiology, Tissue Engineering and Biomaterials Volume, Springer-Verlag Berlin Heidelberg, 2011:p. 69–120.

[11] Goriely, A., Geers, M. G. D., Holzapfel, G. a, Jayamohan, J., Jérusalem, A., Sivaloganathan, S., Squier, W., Dommelen, J. a W. Van, Waters, S., and Kuhl, E., Mechanics of the brain perspectives , challenges , and opportunities. Biomech. Model. Mechanobiol., 2015. 14(5): pp. 931–65.

[12] Ji, S., and Zhao, W., A Pre-computed Brain Response Atlas for Instantaneous Strain Estimation in Contact Sports. Ann. Biomed. Eng., 2015. 43(8): p. 1877–95.

[13] Ji, S., Zhao, W., Ford, J. C., Beckwith, J. G., Bolander, R. P., Greenwald, R. M., Flashman, L. a, Paulsen, K. D., and McAllister, T., Group-wise evaluation and comparison of white matter fiber strain and maximum principal strain in sports-related concussion. J. Neurotrauma, 2015. 32(7): pp. 441–454.

[14] Ji, S., Zhao, W., Li, Z., and McAllister, T. W., Head impact accelerations for brain strain-related responses in contact sports: a model-based investigation. Biomech. Model. Mechanobiol., 2014. 13(5): p. 1121–1136.

[15] Zhao, W., Ruan, S., and Ji, S., Brain pressure responses in translational head impact: a dimensional analysis and a computational study. Biomech. Model. Mechanobiol., 2015. 14(4): pp. 753–766.

[16] Hardy, W. N., Mason, M. J., Foster, C. D., Shah, C. S., Kopacz, J. M., Yang, K. H., King, A. I., Bishop, J., Bey, M., Anderst, W., and Tashman, S., A study of the response of the human cadaver head to impact. Stapp Car Crash J., 2007. 51(October): p. 17–80.

[17] Hardy, W. N., Foster, C. D., Mason, M. J., Yang, K. H., King, a I., and Tashman, S., Investigation of Head Injury Mechanisms Using Neutral Density Technology and High-Speed Biplanar X-ray. Stapp Car Crash J., 2007. 45(November): p. 337–68.

[18] Nahum, A., Smith, R., and Ward, C., Intracranial pressure dynamics during head impact, SAE Tech. Paper, 1977: No. 770922.

[19] Trosseille, X., Tarriere, C., Lavaste, F., Guillon, F., and Domont, A., Development of a FEM of the human head according to a specific test protocol. In Proc. 36th Stapp Car Conf.. 1992: p. 235–253.

[20] Sabet, A. A., Christoforou, E., Zatlin, B., Genin, G. M., and Bayly, P. V, Deformation of the human brain induced by mild angular head acceleration. J. Biomech., 2008. 41(2): p. 307–15.

[21] Bradshaw, D., and Morfey, C., Pressure and shear response in brain injury models. In Proc. 17th Int. Tech. Conf. ESV. 2001: pp. 1–10.

[22] Franklyn, M., Fildes, B., Zhang, L., Yang, K., and Sparke, L., Analysis of finite element models for head injury investigation: reconstruction of four real-world impacts. Stapp Car Crash J., 2005. 49: pp. 1–32.

[23] Takhounts, E. G., Ridella, S. A., Hasija, V., Tannous, R. E., Campbell, J. Q., Malone, D., Danelson, K., Stitzel, J., Rowson, S., and Duma, S., Investigation of traumatic brain injuries using the next generation of simulated injury monitor (SIMon) finite element head model. Stapp Car Crash J., 2008, 52: p. 1–31.

[24] Sternbach, G. L.,2000, The Glasgow Coma Scale. J. Emerg. Med., 2000, 19(1): p. 67–71.

[25] Newman, J. A., Beusenberg, M. C., Shewchenko, N., Withnall, C., and Fournier, E., Verification of biomechanical methods employed in a comprehensive study of mild traumatic brain injury and the effectiveness of American football helmets. J. Biomech., 2005, 38: p. 1469–1481.

[26] Allison, M. a, Kang, Y. S., Bolte, J. H., Maltese, M. R., and Arbogast, K. B., Validation of a helmet-based system to measure head impact biomechanics in ice hockey. Med. Sci. Sports Exerc.,2014. 46(1): p. 115–23.

[27] Pearce, C. W., and Young, P. G., On the Pressure Response in the Brain due to Short Duration Blunt Impacts. PLoS One, 2014. 9(12): No. e114292.

[28] Hoshizaki, B., Post, A., Marshall, K., Clara, K., and Susan, B., The Relationship between Head Impact Characteristics and Brain Trauma, J. Neurol. Neurophysiol, 2013. 5(1): p. 1–81.

[29] Zhao, W., and Ji, S., Parametric investigation of regional brain strain responses via a pre-computed atlas. In IRCOBI Conference. 2015. Lyon, France, No. IRC-15-32.

[30] Zhao, W., and Ji, S., Feasibility of a pressure response superposition scheme in blunt head impact. In IRCOBI Conference. 2015. Lyon, France, pp. 227–228.