A Comparison between two Methods of Head Impact Reconstruction

Arghavan Talebanpour, Lloyd Smith
School of mechanical and material engineering
Washington State University

Abstract

Reconstructing head impacts using computational models is an important tool in understanding brain injury mechanisms. Head impacts are often reconstructed by impacting an Anthropomorphic Test Device (ATD) with an object of interest. The head accelerations are measured and applied to a biofidelic finite element model. Little work has been done, however, to determine how the ATD accelerations applied to a numeric model approximate the brain response an actual impact. The following considered head impacts from a solid sports ball. The brain response of a biofidelic model was compared in two scenarios: accelerations were applied to the model from an impacted by a ball; the head-ball impact was simulated directly in LS-DYNA® with the same speed, direction, and location as occurred with the ATD. Comparison of the linear and the rotational acceleration of the ATD and the biofidelic model showed the inability of the ATD to accurately mimic the dynamic response of the human head in ball impacts. The difference between the peak rotational acceleration and the peak rotational velocities was found to have the highest effect of the brain response of the two accident reconstruction methods.

Introduction

Impacts to the head that lead to the relative movement of the brain with respect to the skull can damage the delicate structure of the brain. Thus, brain injury is one of the most important health concerns in sport [1]. Several studies are dedicated to understand the mechanism of the brain injury and developing safety equipment to protect the brain from impacts [2][3][4][5][6]. Data from actual impacts are either processed statistically [7][8] to find the correlation between the dynamic responses of the head (i.e. linear and rotational accelerations) or used as inputs to computational models to find the brain mechanical responses to the impacts[9][10][11]. Although statistical analysis is a powerful tool in finding a correlation, it doesn`t describe the mechanism of brain injury. Computational models, however, can be used to reconstruct impacts to investigate the mechanical head injury measures such as Maximum Principal Strain (MPS), maximum von Mises stress, and Cumulative Strain Damage Measure (CSDM)[1][12]. Furthermore, sensitivity studies on impact locations or speed can be conducted by computational models to better understand the mechanisms of the brain injury.

Reconstructing head impacts using computational models can be performed in two ways. The most common method is to apply the dynamic responses (accelerations or velocities) of the center of gravity of the head in an impact on the rigid skull of a biofidelic model of the human head [10][9]. Another method is to simulate the actual impact using a biofidelic model [13][14].

Although the dynamic performance of the ATDs and biofidelic head models are verified separately using the cadaveric data [15][16][17], no study has compared them in same impacts. In this work, softball-head impacts at three locations and were reconstructed using an ATD and a biofidelic human finite element (FE) model. The linear and rotational accelerations of the head and the MPS, maximum von Mises stress and the CSDM of the brain in each impact were compared for the two methods.

[1] CSDM measures the volume fraction of the brain experiencing the strain magnitudes higher than a threshold and therefore is associated with the Diffuse Axonal Injury (DAI).

Methods

Experimental setup

An air canon fired softballs (Worth, Gold dot) at 25 m/s, without spin at three locations, as indicated in Fig. 1. The ATD was a 50% male Hybrid III [18] (78051-61X-3623-H, manufacturer) with a 3-2-2 accelerometer array. The head was mounted on a low friction sled which allowed the free movement of the head in the direction of the impact [19]. Each impact was repeated four times and averaged for subsequent analysis. Ball speed was limited to 25 m/s to avoid damaging the ATD. Accelerations were filtered using the 4th order Butterworth filter with a cutoff frequency of 1kHz for linear acceleration and 300 Hz for rotational acceleration [20].

.

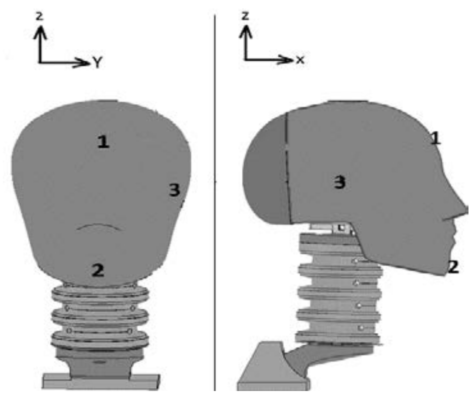

Fig.1 Ball impact locations 1: forehead 2: chin, 3: side.

Impact Reconstruction Models

A 50th percentile male THUMS model version 4.02[21] was used as the biofidelic model in this study. The simulations were performed using LS-DYNA (version mmps R8.1.1). To reconstruct the impacts from the ATD responses, the average linear and rotational accelerations of the head were applied as prescribed motions to the skull. To allow an acceleration input, the skull properties were changed to rigid and constrained to move as a single rigid part, as done elsewhere [9][10][11]. In the following, this reconstruction method will be referred to as “accelerated”.

The softball-head impacts were also reconstructed by simulating a softball impact with the biofidelic model directly [22], hereafter referred to as “impacted”. The ball was modeled from one homogeneous rate dependent material using LS-DYNA MAT 57 [22]. An initial velocity was applied to the nodes of the ball, and the biofidelic model (with a compliant skull) was free to recoil after impact. The contact between the head and the ball was accomplished using Automatic Surface to Surface Contact with a coefficient of friction of 0.1. The head acceleration was obtained from the derivative of the head velocity using a novel MATLAB script that also eliminates noise [23].

Results

The temporal linear acceleration and rotational acceleration of the biofidelic FE model and the ATD at 25 m/s for three impact locations are shown in Fig. 2.

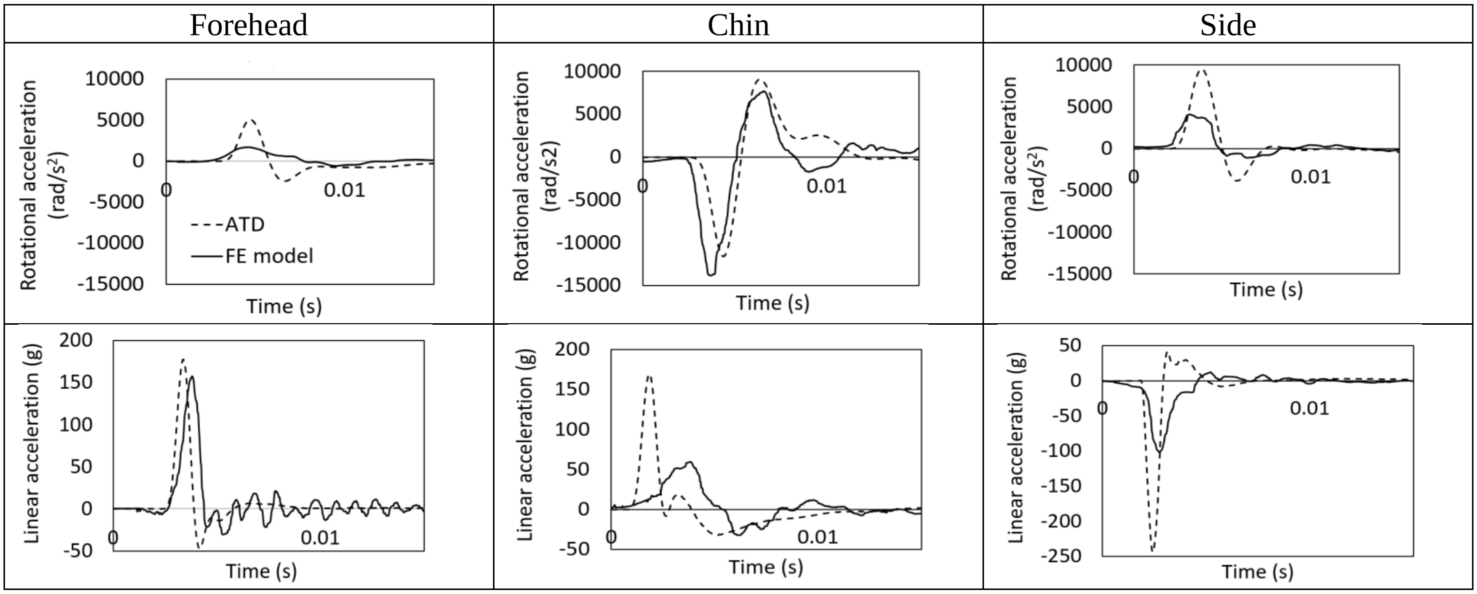

Fig.2 Accelerations of ATD and biofidelic FE model from a 25 m/s softball impact.

Table 1 compares the dynamic responses (velocity and acceleration) and the brain injury measures (MPS, maximum von Mises stress, and CSDM) of the impacted and accelerated models.

Impact location	Impact speed (m/s)	Model	Peak Linear Acceleration (g)	Peak Rotational Acceleration (rad/s^2)	Peak Rotational velocity (rad/s)	Max von Mises Stress (kPa)	Max Principal Strain	CSDM (0.08)
Forehead	25	Impacted	161	1658	5.00	2.66	0.14	0.0012
Forehead	25	Accelerated	177	4946	6.50	2.16	0.12	0.0010
Chin	25	Impacted	58	13771	24.00	6.37	0.36	0.2000
Chin	25	Accelerated	167	11529	15.00	5.58	0.32	0.0670
Side	25	Impacted	104	3993	7.70	2.20	0.14	0.0020
Side	25	Accelerated	244	9350	13.50	3.43	0.21	0.0380

Table.1 Dynamic and biofidelic comparison of the accelerated and impacted models.

Discussion

In this work, softball-head impacts were reconstructed using a biofidelic FE model. The accelerated model used experimental accelerations from a ATD-softball impact, while the impacted model simulated the ball impact directly. The difference between the movement of the accelerated and the impacted models resulted in up to a 65% difference in the CSDM. Neither the accelerations nor the CSDM of one method was consistently higher than the other, making their comparison case dependent. In the forehead impact, the accelerated model primarily experiences translation with little neck constraint. The impacted model, on the other hand, shows rotation and movement of the neck from a forehead impact. The neck from a chin impact of the accelerated model had a larger contribution than the forehead impact, resulting in higher rotational acceleration. The results show that the accelerations of the accelerated model are closer to the impacted model for frontal impacts (forehead and chin) than the side impacts. This is not surprising, since the Hybrid III neck was not designed for side impacts [24].

The impact location on the head (ie. forehead, chin or side) significantly affected the brain deformation. Table 1 shows that the chin impact had the highest and the forehead impact had the lowest brain injury measures in both the accelerated and impacted models. Note that both the peak rotational acceleration and velocity is highest in the chin and lowest in the forehead, and the linear acceleration of the chin impact is lower than the forehead. This suggest that brain injury measures have stronger dependence on rotational kinematics than linear kinematics.

The peak rotational acceleration and the impact duration both affect the severity of the impact [25][26][3]. As for the chin impacts, the difference between the rotational acceleration of the accelerated and impacted models are not significant (within 16%) but the difference between the CSDMs are (within 66%) which may be due to the difference in peak rotational velocity (40%). This suggests that the longer the head is experiencing the rotational acceleration, the fraction of the brain experiencing large deformation increases. The forehead impacts had relatively low rotational acceleration, resulting in low injurious accelerations, and a reduced dependence on duration.

Linear acceleration of the head might not be highly related to the brain deformation as the impacted linear accelerations are roughly one third of accelerated linear accelerations for the chin impacts, yet the brain injury measures of the impacted are higher than the accelerated model.

Limitations

The difference between the accelerations of the ATDs and the biofidelic head models is the most significant reason for the different brain deformations predicted by the two accident reconstruction methods. However, any other dissimilarity between these two methods (i.e. the rigid skull for the accelerated THUMS simulations, and the accuracy of modeling impact conditions) can affect the comparison of their resulting brain injury measures. In this study, only one type of impact (softball-head) was reconstructed as an example of how different the two accident reconstruction methods can be. It is not known how other types of impact scenarios will compare. Furthermore, in this study only one ATD (Hybrid III) and one biofidelic FE model (THUMS) were used. A low softball speed (25 m/s) was used here to avoid the damage of the Hybrid III ATD. This resulted in brain deformation less than the injurious levels proposed by other studies [27][9] [28].

Conclusion

The dynamic and brain injury measures of two accident reconstruction methods were compared in three softball-head impacts. The significant differences between the linear and rotational accelerations were the main reason for the difference in CSDM predicted by the two methods. The inability of the accelerated model to describe the impacted model motions may lead to inaccurate brain deformation predictions. Although the peak rotational acceleration of the head was correlated with the maximum stress and strain of the brain elements, the fraction of the brain elements that pass a strain threshold (CSDM) was affected by the rotational velocity. Linear acceleration was found to not be significantly correlated with the brain responses.

References

[1] A. A. Hyder, C. A. Wunderlich, P. Puvanachandra, G. Gururaj, and O. C. Kobusingye, "The impact of traumatic brain injuries: A global perspective," *NeuroRehabilitation*, vol. 22, pp. 341–353, 2007.

[2] T. El Sayed, A. Mota, F. Fraternali, and M. Ortiz, "Biomechanics of traumatic brain injury," *Comput. Methods Appl. Mech. Eng.*, vol. 197, no. 51–52, pp. 4692–4701, 2008.

[3] E. G. Takhounts, S. a Ridella, S. Rowson, and S. M. Duma, "Kinematic Rotational Brain Injury Criterion (BRIC)," *22nd Enhanc. Saf. Veh. Conf. Proc.*, p. Paper No. 11-0263, 2011.

[4] F. A. Fernandes and R. J. A. de Sousa, "Head injury predictors in sports trauma – A state-of-the-art review," *Proc. Inst. Mech. Eng. Part H J. Eng. Med.*, vol. 229, no. 8, pp. 592–608, 2015.

[5] W. J. Curnow, "The efficacy of bicycle helmets against brain injury," *Accid. Anal. Prev.*, vol. 35, no. 2, pp. 287–292, 2003.

[6] A. Post, A. Oeur, B. Hoshizaki, and M. D. Gilchrist, "An examination of American football helmets using brain deformation metrics associated with concussion," *Mater. Des.*, vol. 45, pp. 653–662, 2013.

[7] R. M. Greenwald, J. T. Gwin, J. J. Chu, and J. J. Crisco, "Head impact severity measures for evaluating mild traumatic brain injury risk exposure," *Neurosurgery*, vol. 62, no. 4, pp. 789–798, 2008.

[8] S. Rowson *et al.*, "Rotational head kinematics in football impacts: An injury risk function for concussion," *Ann. Biomed. Eng.*, vol. 40, no. 1, pp. 1–13, 2012.

[9] S. Kleiven, "Predictors for traumatic brain injuries evaluated through accident reconstructions," *Stapp Car Crash J.*, vol. 51, pp. 81–114, 2007.

[10] E. G. Takhounts *et al.*, "Investigation of traumatic brain injuries using the next generation of simulated injury monitor (SIMon) finite element head model," *Stapp Car Crash J.*, vol. 52, no. November, pp. 1–31, 2008.

[11] M. C. Doorly and M. D. Gilchrist, "The use of accident reconstruction for the analysis of traumatic brain injury due to head impacts arising from falls.," *Comput. Methods Biomech. Biomed. Engin.*, vol. 9, no. 6, pp. 371–377, 2006.

[12] F. . Bandak, A. . Zhang, R. . Tannous, F. DiMasi, P. Masiello, and R. Eppinger, "SIMon: A SIMULATED INJURY MONITOR; APPLICATION TO HEAD INJURY ASSESSMENT," pp. 1–7, 1990.

[13] M. Fahlstedt, P. Halldin, and S. Kleiven, "The protective effect of a helmet in three bicycle accidents - A finite element study," *Accid. Anal. Prev.*, vol. 91, pp. 135–143, 2016.

[14] M. Ghajari, S. Peldschus, U. Galvanetto, and L. Iannucci, "Effects of the presence of the body in helmet oblique impacts," *Accid. Anal. Prev.*, vol. 50, pp. 263–271, 2013.

[15] B. R. Cobb, A. MacAlister, T. J. Young, A. R. Kemper, S. Rowson, and S. M. Duma, "Quantitative comparison of Hybrid III and National Operating Committee on Standards for Athletic Equipment headform shape characteristics and implications on football helmet fit," *Proc. Inst. Mech. Eng. Part P J. Sport. Eng. Technol.*, vol. 229, no. 1, pp. 39–46, 2015.

[16] A. Kim *et al.*, "A Comparison of the BioRID II, Hybrid III, and RID2 in Low-Severity Rear Impacts," *19th Int. Tech. Conf. Enhanc. Saf. Veh.*, pp. 1–27, 2005.

[17] A. Bartsch, E. Benzel, V. Miele, D. Morr, and V. Prakash, "Hybrid III anthropomorphic test device (ATD) response to head impacts and potential implications for athletic headgear testing," *Accid. Anal. Prev.*, vol. 48, pp. 285–291, 2012.

[18] S. A. Hassan and E. Al., "Hybrid III—ABiomechanicallyBased Crash Test Dummy," *Entomophaga*, vol. 39, no. 1, pp. 107–119, 1994.

[19] N. O. Committee, O. N. Standards, and F. O. R. Athletic, "STANDARD PROJECTILE IMPACT TEST METHOD AND EQUIPMENT USED IN EVALUATING THE PERFORMANCE CHARACTERISTICS OF PROTECTIVE HEADGEAR , FACEGUARDS OR PROJECTILES NOCSAE DOC (ND) 021- 12m13 NATIONAL OPERATING COMMITTEE," no. June 2012, 2014.

[20] "SAE J211-1 (1995): Instrumentation for Impact Test, Part 1, Electronic Instrumentation," 2010.

[21] M. Iwamoto, Y. Nakahira, A. Tamura, and M. Iwamoto, "DEVELOPMENT OF ADVANCED HUMAN MODELS IN

THUMS AUTHORS : CORRESPONDENCE : KEYWORDS : THUMS-AM50," pp. 47–56.

[22] S. D. Burbank and L. V Smith, "Dynamic characterization of rigid foam used in finite element sports ball simulations," *Proc. Inst. Mech. Eng. Part P J. Sport. Eng. Technol.*, vol. 226, no. 2, pp. 77–85, 2012.

[23] R. Chartrand, "Numerical Differentiation of Noisy, Nonsmooth Data," *ISRN Appl. Math.*, vol. 2011, pp. 1–11, 2011.

[24] N. Yoganandan, F. A. Pintar, D. J. Maiman, M. Philippens, and J. Wismans, "Neck forces and moments and head accelerations in side impact," *Traffic Inj. Prev.*, vol. 10, no. 1, pp. 51–57, 2009.

[25] H. Kimpara and M. Iwamoto, "Mild traumatic brain injury predictors based on angular accelerations during impacts," *Ann. Biomed. Eng.*, 2012.

[26] L. Ren, D. Baumgartner, J. Yang, J. Davidsson, and R. Willinger, "Investigation of diffuse axonal injury induced by rotational acceleration via numerical reconstructions of in vivo rat head impact experiments," *Int. J. Crashworthiness*, vol. 20, no. 6, pp. 602–612, 2015.

[27] C. Deck, D. Marjoux, D. Baumgartner, R. Willinger, and I. De Mécanique, "Head Injury Predictive Tools for Protective Systems Optimisation," pp. 1–13, 2000.

[28] E. G. Takhounts, S. a Ridella, S. Rowson, and S. M. Duma, "Kinematic Rotational Brain Injury Criterion (BRIC)," *22nd Enhanc. Saf. Veh. Conf. Proc.*, p. Paper No. 11-0263, 2011.