行政院國家科學委員會專題研究計畫　成果報告

墜落事故之現場重建及頭部傷害之研究

計畫類別：個別型計畫
計畫編號：NSC94-2213-E-041-002-
執行期間：94年08月01日至95年10月31日
執行單位：嘉南藥理科技大學嬰幼兒保育系

計畫主持人：張立東
共同主持人：蔡明哲
計畫參與人員：C. D. Huber, K-H. Yang

報告類型：精簡報告
報告附件：出席國際會議研究心得報告及發表論文
處理方式：本計畫可公開查詢

中　華　民　國　95年10月12日

行政院國家科學委員會補助專題研究計畫 ■ **成 果 報 告** □**期中進度報告**

墜落事故之現場重建及頭部傷害之研究

計畫類別：■ 個別型計畫　　□ 整合型計畫
計畫編號：NSC 94－2213－E －041 －002－
執行期間：　94 年　8 月　1 日 至　95 年　10 月　31 日

計畫主持人：張立東
共同主持人：蔡明哲
計畫參與人員： C. D. Huber, K-H. Yang

成果報告類型(依經費核定清單規定繳交)：■精簡報告　□完整報告

本成果報告包括以下應繳交之附件：
□赴國外出差或研習心得報告一份
□赴大陸地區出差或研習心得報告一份
■出席國際學術會議心得報告及發表之論文各一份
□國際合作研究計畫國外研究報告書一份

處理方式：除產學合作研究計畫、提升產業技術及人才培育研究計畫、列管計畫及下列情形者外，得立即公開查詢
□涉及專利或其他智慧財產權，■一年□二年後可公開查詢

執行單位：嘉南藥理科技大學

中　華　民　國　95　年　10　月　11　日

THE STUDY OF FALL ACCIDENT RECONSTRUCTION AND HEAD INJURY
墜落事故之現場重建及頭部傷害之研究

L-T. Chang[1], C. D. Huber[2], M-C. Tsai[3], K-H. Yang[2]
[1]Dept. of Childhood Education and Nursery, Chia Nan University of Pharmacy and Science
[2]Dept. of Biomedical Engineering, Wayne State University, Detroit, Michigan, USA
[3]Dept. of Emergency Medicine, National Chung Kung University Hospital
(Grant No.: NSC 94-2213-E-041-002)

1. Abstract

This study endeavors to investigate the kinematics and head accelerations of real-world pediatric headfirst falls in order to improve understanding of how head and brain injuries occur in children. The mechanisms and thresholds of pediatric head injury are not well understood, mainly due to difficulties in obtaining pediatric specimens. The results from this set of new data using multi-body modeling should serve to provide researchers a chance to investigate *in vivo* pediatric head injury and reiterate the need for more biofidelic models and pediatric post-mortem human subject testing.

Keywords: children, biomechanics, multi-body model

中文摘要

本研究根據臨床兒童墜落案例，分析兒童墜落時之運動學過程及撞擊時頭部承受之加速度值，進而探討兒童頭部傷害之機制及準據。結果顯示本研究建立之多體動力模型具有良好之生物相似性及實驗重複性，可作為研究人員探討兒童頭部傷害之工具。

關鍵詞：兒童、生物力學、多體動力模型

2. Introduction

Pediatric falls are a leading cause of accidental death in children. Falls result in more emergency room visits and hospital admissions than any other source of injury [1,2]. Head trauma, specifically, is the most common injury in children. It has been reported that in the United States alone, 500,000 emergency room visits and 95,000 hospital admissions can be attributed annually to pediatric head injuries, at a cost of $1 billion [3,4]. Despite the prevalence of pediatric head injury, the mechanisms of head injury are not as well understood in children as in adults. It is clear that the mechanisms of the same injury can vary as a function of age, due to the differences in the developing anatomy of children. Of particular interest in head trauma is the development of the skull, brain, neck muscles and cervical spine.

It is well-known that children are not simply small adults, but rather underdeveloped adults. However, current pediatric testing surrogates are designed based on data scaled from adults [5,6]. Because of the aforementioned nature of physiologic development in children, scaling may not accurately represent the properties of the pediatric body. Injury tolerance curves for the pediatric population are based mainly on scaling from adult cadaveric impact responses

The lack of knowledge in the area of pediatric injury response is due to several factors, most notably the difficulty in procuring pediatric post-mortem human subjects. Because of this, methods must be developed in order to study pediatric head injury without using specimens. Mathematical modeling has been used successfully to study adult head and brain injury, but many of the more sophisticated finite element models are still based on post-mortem human subject data for model validation. Since this data is unavailable for children, an intermediate modeling step may be useful. Multi-body models can provide kinematics and biomechanical data, which can yield insight into injury if the models are based on well-characterized known injury scenarios.

3. Methods

Data collection

The cases used in this study were collected at the National Cheng Kung University Hospital in Tainan, Taiwan during a period from December 2000 to December 2003. Each subject sustained a head injury related to a headfirst fall and was admitted to the emergency room. This study utilizes medical diagnosis and data available from physicians, as well as additional data collected by field investigators. These investigators were sent to the scene of the fall to gather firsthand accounts from eyewitnesses and to survey the scene in detail. The circumstances surrounding the fall, head and body orientations of the subject at impact, fall height and impact surface were all recorded. Data was collected for eight subjects with ages ranging from 0.2 to 7.5 years that fit the criteria for this study. The respective age, weight and height are shown in Table 1.

Model generation

To study the kinematics and biomechanics of the falls, the TNO P-series child dummy multi body models developed for MADYMO were used for numerical simulations. In order to most realistically simulate the cases, the most appropriately-aged child dummy model was chosen for each case as shown in Table 1. Because the P0 TNO child dummy was not available, only the oldest seven of the eight subjects were modeled.

4. Results

To simulate the kinematics of the fall, initial linear and rotational velocities were assumed in such a way that the impact occurred in a similar manner as described by the physicians and eyewitnesses, as shown in Table 2. The impact positions predicted by the models are shown in Figure 1.

The P-series child dummies have a seated default position. For some cases, the joints were adjusted in order to position the dummy in a standing posture or in an initial position based on the case descriptions, as shown in Table 3. Prescribed initial velocities were in accordance with case descriptions while marinating the same impact site as described, but kept as small as possible in order to minimize errors. Impact surfaces were modeled as rigid planes because the impact surfaces in the real world cases were overwhelmingly rigid materials, such as concrete, stone and asphalt. The only exception was Case 5, in which the impact surface was dirt. It is believed that the rigid plane surface used in the models reasonably approximates such a surface and will not adversely affect the results. The friction coefficients were based on work done by Bertocci, et al. [7]

Table 4 shows the impact severity based on the maximum resultant acceleration, clinical measurements such as the Glasgow coma scale (GCS), abbreviated injury scale (AIS), and the Overall Head Injury Measure (OHIM) of the injury severity, and clinical diagnosis. Please note that a GCS of 15 indicates no or minor impairment and a decreasing score indicates greater impairment, while an AIS or OHIM of 1 indicates minor injury and larger scores indicate increasing severity. The accelerations calculated are linear only, since this series of MADYMO models did not calculate angular accelerations.

5. Discussion

The 15-segement TNO P-series child dummy MADYMO models were chosen for its availability. The original intent was to modify the inertial and joint properties using patient-specific information and information from the Generator of Body Data (GEBOD) to calculate more anthropometrically correct model for simulations. However, encryption of some joint characteristics did not allow for changes. Consequently, the closest size dummy model was used for this study. Unfortunately, model calculated accelerations at the head's center of gravity were unrealistically high and did not seem to accurately represent the real world fall. There are several possible reasons for this. Firstly, these models clearly do not incorporate active muscle involvement that occurs when a human being is conscious, which is a grave disadvantage when studying falls. It is possible that the subjects tried to "catch" themselves with an outstretched arm or bent at the neck in anticipation of the impact. This would have obviously affected the results of the real-world falls, but cannot be accurately simulated in current modeling efforts. Secondly, the initial velocities used to force the dummies to land appropriately may not be representative of the actual initial conditions of the real world case. The data collected lacked detail on this issue, and this could have affected the results. Lastly, these dummy models have not been validated for fall scenarios. The P-Series child dummy models were validated against experimental data for frontal impact. In addition, the P6 model was validated for cases of lateral loading. Although the models predicted kinematics and accelerations well in these validation tests, there is no guarantee that the models will perform well in the case of headfirst falls with complicated kinematics. Although it is possible to conduct a physical dummy experiment to determine if the problem lies with the model or the dummy itself, there is still no real world data to validate the results obtained from dummy experiments. Consequently, the experiment was not considered.

It has been reported by O'Riordain et al. [8] that the resulting accelerations were too high when using the default MADYMO contact force characteristics in reconstruction of fall cases. It was speculated that this unrealistic result came from the use of aluminum headforms to create the default curves instead of post-mortem human subjects, which would yield a more biofidelic response. Unfortunately, insignificant changes in head acceleration were found after changing to the same head contact force-deflection curve as used in the O'Riordain study. Note that the contact characteristics of a child's head on impact have not been well-studied for all pediatric age groups, limiting the utility of using alternate curves.

The comparison between the predicted relative impact severity and the clinical injury severity does allow one to draw some insight from the models of these case studies. The only fatal impact modeled, Case 7, did not result in the largest predicted head accelerations. Instead, two of the cases modeled using the P3/2 dummy gave higher results, although the injuries in those cases were not as severe according to clinical data. Case 4 also showed very high head acceleration and was modeled with the P3/2 dummy. It is possible the high values found for Cases 1, 3 and 4 are due to differences in the P3/2 dummy as compared to the other P-series dummies. However, the differences in the P3/2 dummy are not related to the head and neck, but to the thorax, which may affect the kinematics of the neck. Results from models using other P-series dummies seemed to be more reasonable.

6. Conclusions

Although the absolute data generated by these models does not accurately predict the accelerations seen in the real world cases, it may allow for comparison of the relative impact severity between cases with their relative head injury severity if the same dummy model is used. Even with well-characterized cases, the current lack of data involving child impact response impedes the ability of researchers to create biofidelic models. Results from this study could be greatly improved with the development of a biofidelic child model that could be validated against crash impacts and fall impacts for maximum utility. In order to develop improved test dummies or dummy models, more data is needed on pediatric response in impact situations. This data could also be used to develop improved contact characteristics for pediatric multi body models or to develop sophisticated finite element models which are a more economical alternative to dummy tests.

7. Reference

[1] Gallagher, S., et al., The incidence of injuries among 87,000 Massachusetts children and adolescents: results of the 1980-81 Statewide Childhood Injury Prevention Program Surveillance System. Am J Public Health, 1984. 74(12): p. 1340-1347.

[2] Pillai, S.B., et al., Fall injuries in the pediatric population: safer and most cost-effective management. Journal of Trauma, 2000. 48(6): p. 1048-1051.

[3] Krauss, J., A. Rock, and P. Hemyari, Brain injuries among infants, children, adolescents, and young adults. Am J Dis Child, 1990. 144: p. 684-691.

[4] Rivara, F., Childhood injuries III: epidemiology of non-motor vehicle head trauma. Dev Med Child Neurol, 1984. 26: p. 81-87.

[5] Irwin, A. and H. Mertz, Biomechanical basis for the CRABI and Hybrid III child dummies. 41st Stapp Conference, SAE Paper #973317, 1997. p. 1-12.

[6] Melvin, J.W. Injury assessment reference values for the CRABI 6-moth infant dummy in arear-facing infant restraint with airbag deployment. in SAE International Conference and Exposition. 1995. Detroit, MI.

[7] Bertocci, G.E., et al., Influence of fall height and impact surface on biomechanics of feet-first free falls in children. Injury Int J Care Injured, 2004. 35: p. 417-424.

[8] O' Riordain, K., et al. Reconstruction of real world head injury accidents resulting from falls using multibody dynamics. Clin Biomech, 2003. 18: p. 590-600.

Table 1. Age and weight compared with dummy age chosen

Case	Age (years)	Weight (kg)	P Series Dummy
1	1.2	9.2	P3/2
2	0.2	6.0	NA
3	1.5	11.5	P3/2
4	1.0	8.0	P3/2
5	7.4	28.0	P6
6	7.5	25.0	P6
7	3.0	20.0	P3
8	2.9	14.0	P3

Table 2. Kinematic case descriptions

Case	Fall Height (m)	Fall Direction	Head Injury Location
1	2.2^f	Forward	Right occipital
2	7.7^h	Forward	Left parietal
3	3.5^h	Forward	Left frontal
4	0.8^h	Backward	Middle occipital
5	2.6^h	Backward	Left temporal
6	0.65^h	Downward	Upper parietal
7	6.2^f	Forward	Right frontal
8	1.1^h	Forward	Right occipital

Table 3. Initial positions and velocities for each model

Case	Hip Joint (rotation about y-axis)	Knee Joint (rotation about y-axis)	Orientation (forward rotation)	Forward Velocity	Angular velocity (x-rot)	Angular velocity (y-rot)	Angular velocity (z-rot)
1	1.5	-1.5	0.0	0.1	0.0	2.8	4.5
3	0.0	0.0	0.0	0.4	0.0	2.2	0.0
4	0.0	0.0	-0.5	-0.1	0.0	-10	0.0
5	1.5	-1.5	0.0	-0.1	-1.8	-1	0.0
6	1.5	-1.5	3.14	0.5	0.0	0.0	0.0
7	1.5	-1.5	0.0	0.1	0.0	14.5	0.0
8	1.5	-1.5	0.0	0.1	0.0	5	7

Table 4. Predicted impact severity compared with clinical injury severity

Case	Max. Resultant (G), in Descending Order	GCS	Max. Head AIS	OHIM	Head Injury
3	1018	7	4	4.5	Brain contusion, intracranial hemorrhage
1	980	15	1	1	Subgalea hematoma
7	670	6	4	5.4	Fatal; multiple skull fractures, brain contusion, subarachnoid and intercranial hemorrhages
4	523	14	3	3	Head contusion
5	185	15	1	1	Head laceration
6	135	15	3	3	Scalp hematoma, mild concussion, brain swelling
8	80	15	1	1	Scalp hematoma

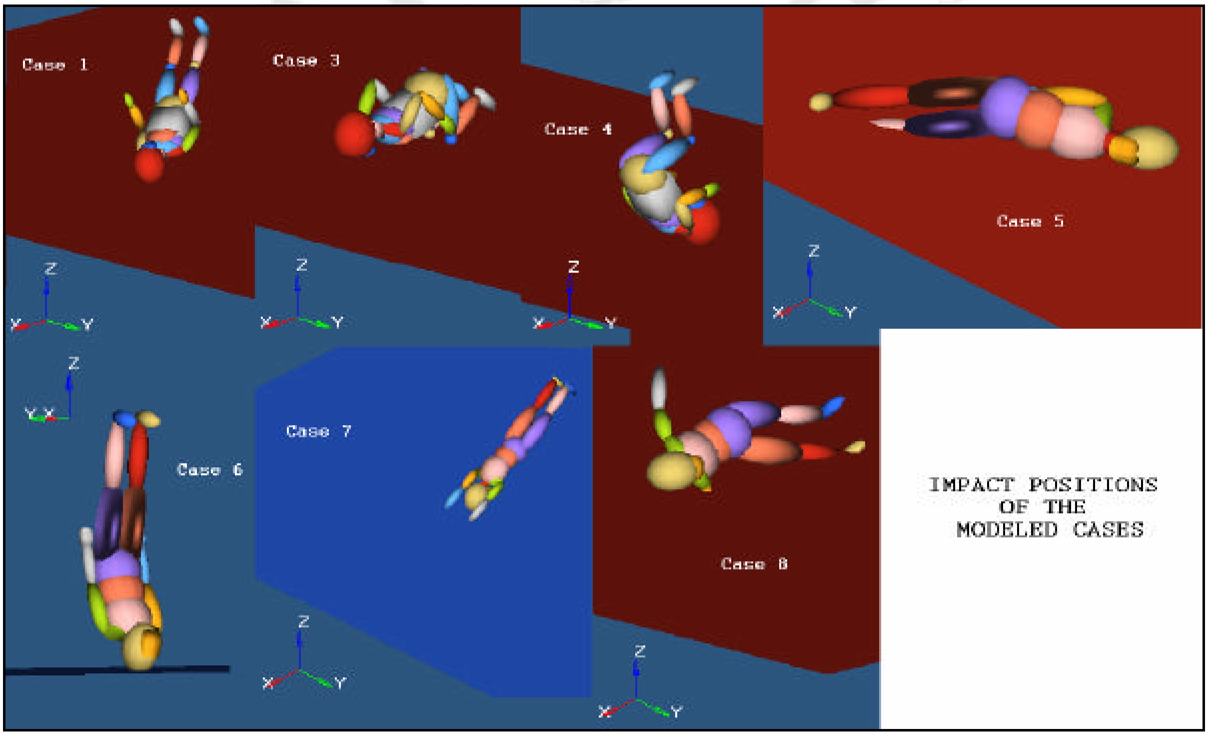

Figure 1. Head impact locations of the MADYMO models