The Tolerance and Response of the Frontal Bone to Blunt Impact

J. Cormier[1], J. Bisplinghoff[1], S. Manoogian[1], S. Rowson[1], A. Santago[1], C. McNally[1], J. Bolte[2] and S. Duma[1]

[1] Virginia Tech Center for Injury Biomechanics
[2] The Ohio State University; Transportation Research Center

ABSTRACT

The current understanding of the response of the frontal bone to blunt impact is limited. Previous studies have utilized vastly different methods which prevents statistical analyses to determine the tolerance of the frontal bone. The purpose of this study is to determine the tolerance of the frontal bone to blunt impact and to define corridors describing its force-displacement response. The response corridors were used to evaluate the biofidelity of the Facial and Ocular CountermeasUre Safety (FOCUS) headform, a sophisticated headform capable of measuring forces imposed on eight separate facial regions. Risk functions for fracture were developed using parametric and non-parametric techniques suggesting that 2,500 N represents a 50% risk of fracture. The use of acoustic emission is essential due to the increase in impactor force after fracture onset. The force-displacement response of the frontal bone was found to vary by the type of fracture sustained by the subject. Subjects that were found to have a frontal sinus present within the impacted region had a statistically higher risk of sustaining a fracture. The response of the FOCUS headform was within the force-displacement corridors defined for the cadaveric response.

INTRODUCTION

The previous work can be divided into two main categories by the type of impactor used to strike the face. In one, a cylindrical impactor strikes the subject with the end, consisting of a flat face (Nahum, Gatts et al. 1968; Schneider and Nahum 1972; Nahum 1975) in the other, a cylindrical impactor strikes the subject side-on (Hodgson, Lange et al. 1965; Nyquist, Cavanaugh et al. 1986; Allsop, Warner et al. 1988; Yoganandan, Fintar et al. 1988; Bermond, Kallieris et al. 1989; Cesari, Ramet et al. 1989; Welbourne, Ramet et al. 1989; Yoganandan, Pintar et al. 1993; Bruyere, Bermond et al. 2000) meant to represent a steering wheel impact. In addition to the use of different impactor shapes, the methods of these studies include the use of different impact location and direction and the use of padding. The large variation between and within these studies limits the capacity for statistical methods to produce accurate estimates of fracture risk or tolerance. Testing by Allsop *et al.* utilizing a cylindrical impactor demonstrated that facial bones are capable of supporting load after fracture has initiated (Allsop, Warner et al. 1988; Allsop and Kennett 2002). This is an important concept, as it requires that the fracture force be determined separate from a force-time or force-deflection response which was neglected in all other facial impact studies.

The current knowledge is also lacking in an understanding of the mechanical response of the frontal bone to impact. The development of finite element models or Anthropomorphic Test Devices (ATD) for fracture prediction is limited by the lack of a quantitative measure of frontal bone response to impact. The Facial and Ocular CountermeasUre Safety (FOCUS) headform (FOCUS) headform has the ability to measure forces applied to the facial and orbital structures. There are eight separate sensing regions within the facial structures that overlie embedded load cells (Figure 1).

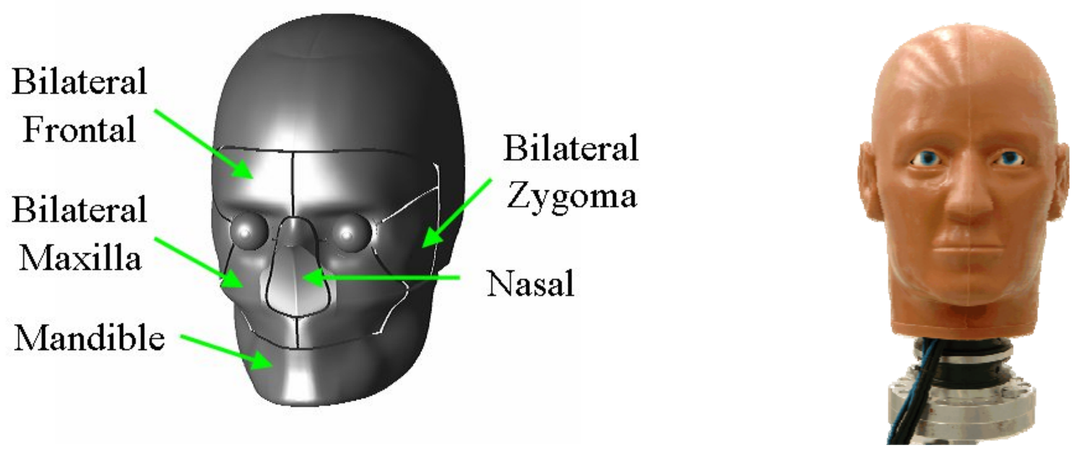

Figure 1: Load cell locations in FOCUS headform (left) and photograph (right).

The first goal of this study is to generate additional data in order to determine risk functions for the prediction of frontal bone fracture as a result of blunt impact. The second goal of this study is to develop an understanding of the cadaveric response to blunt impact and compare this response to the FOCUS headform.

METHODS

The data for this study were obtained by performing facial impacts on male cadaveric subjects. The methods of this study consist of striking the frontal bone with the flat face of an unpadded, cylindrical impactor, along with the use of acoustic emission sensors to determine the time of fracture onset. A total of 28 male subjects ranging in age from 43 to 76 years were included in the study. Pre-test CT imaging was performed on twelve subjects and post-test CT imaging was performed on three specimens. The CT images were used to measure the thickness of the frontal bone and overlying skin at the horizontal level of the impact location. Prior to testing, each head was rigidly mounted to a semi-circular, polycarbonate support using Bondo (Figure 2). Each impact was performed using a cylindrical, free-falling rigid aluminum impactor (3.2 kg) with a steel tip. The flat impacting surface had an area of 6.45 cm^2 (1 in^2) and was machined with a slight bevel to reduce edge effects. An AE sensor (Micro30S, Physical Instruments, New Jersey) was mounted to the frontal bone, just posterior to the apex of the forehead. The attachment method has been used successfully in previous testing by others (Funk, Crandall et al. 2002; Rudd, Crandall et al. 2004; Kent, Stacey et al. 2008). The AE sensors were sampled at a rate of 2 or 5 MHz and all other instrumentation at 30 kHz. In all tests, the rigid impactor was instrumented with two single-axis accelerometers (Endevco 7264B-2000, Endevco Corp., San Juan Capistrano CA). All data except AE data were filtered to CFC 300. The force-displacement response corridor of the cadaver and FOCUS responses were characterized by the mean and standard deviation of the characteristic average (Lessley, Crandall et al. 2004).

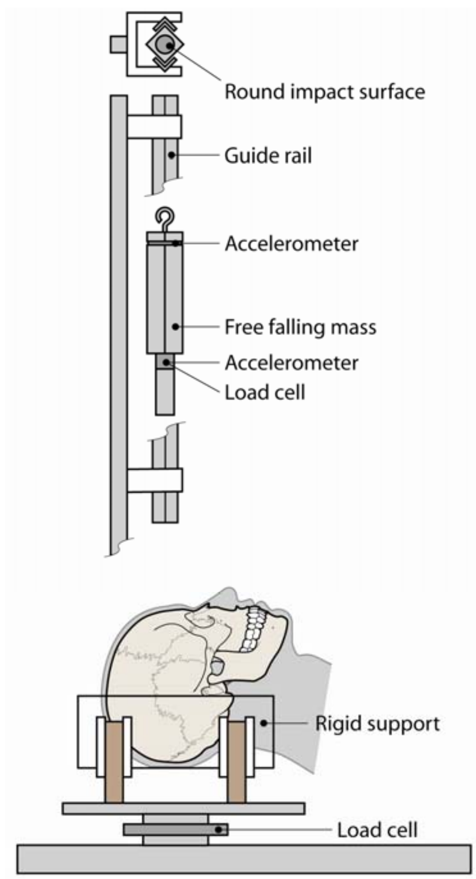

Figure 2: Schematic of test apparatus used in the current study.

AE data were used to determine the force at which fracture initiated. Previous Studies using AE data have denoted the onset of AE signal as the onset of fracture (Allsop, Warner et al. 1988; Funk, Crandall et al. 2002; Funk, Srinivasan et al. 2002). Recently, studies have utilized a threshold to differentiate AE consistent with fracture from a baseline AE signal (Rudd, Crandall et al. 2004; Cormier, Manoogian et al. 2008; Kent, Stacey et al. 2008). Kent *et al.* (2008b) demonstrated that an AE burst corresponded with a sharp decline in force during phalange fracture. The threshold for this study was determined by comparing the AE measured during fracture and non-fracture tests.

The non-censored AE data were utilized in a survival analysis to estimate the relationship between impactor force and fracture risk. A Weibull model was assumed and fit to the data which contained fracture and non-fracture observations. The advantage of using a Weibull model is that the method used to determine the model parameters accounts for the fact that the non-fracture tests are right censored (Allison 1995; Cantor 2003). A non-parametric model was also created using the Kaplan-Meier method. The Kaplan-Meir method assumes the data are only right or non-censored and determines the risk of fracture based on the number of subjects at risk which sustain a fracture for a given force (Kleinbaum and Klein 2005).

RESULTS

A total of 46 tests were performed to determine the tolerance of the frontal bone to blunt impact. The peak force during each impact ranged from 520 to 6400 N. Pulse duration of each impact ranged from 5 to 10 ms. A facial fracture was produced in 19 tests subjects and 11 tests were performed on regions with pre-existing fractures. An Acoustic Emission (AE) signal was measured in every test using a sensor mounted on the frontal bone. A threshold voltage of 9 volts was established for series one, 5 volts for series two and 2 volts for the third series. These thresholds were based on the magnitude of AE during fracture and non-fracture tests (Figure 3).

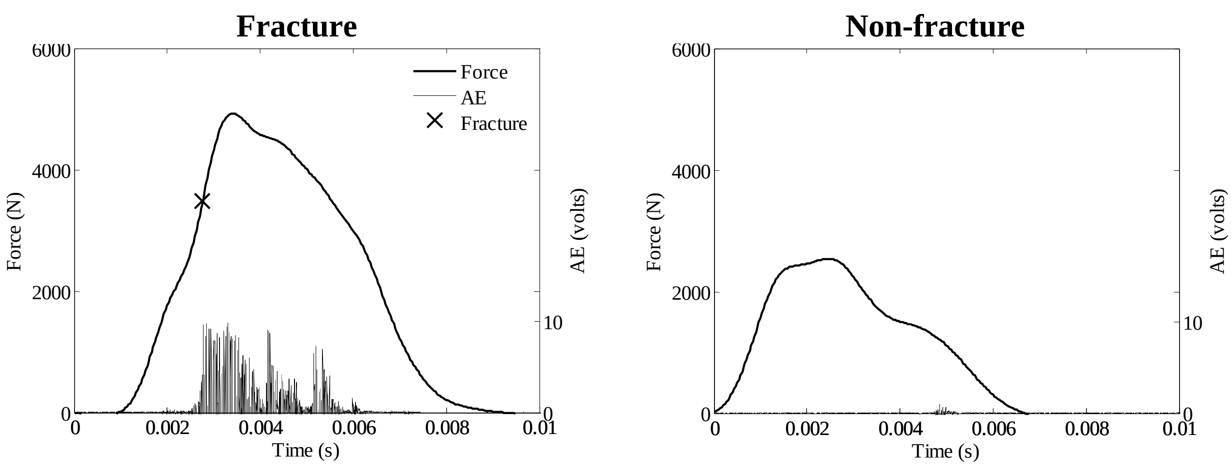

Figure 3: Acoustic emission and force during frontal bone impact.

The average frontal bone thickness was 0.5 cm (Std. Dev. = 3 mm) and the average skin thickness was 0.5 cm (SD = 1 mm). Frontal bone thickness was statistically related to head depth (r^2=0.81, p < 0.0001) and head width (r^2=0.80, p < 0.0001). No correlation was observed between skull and skin thickness and fracture force or any stiffness measure (toe region length, initial and secondary stiffness). Subject age did not have a statistically significant influence on the force at fracture onset (p = 0.96) (Figure 4).

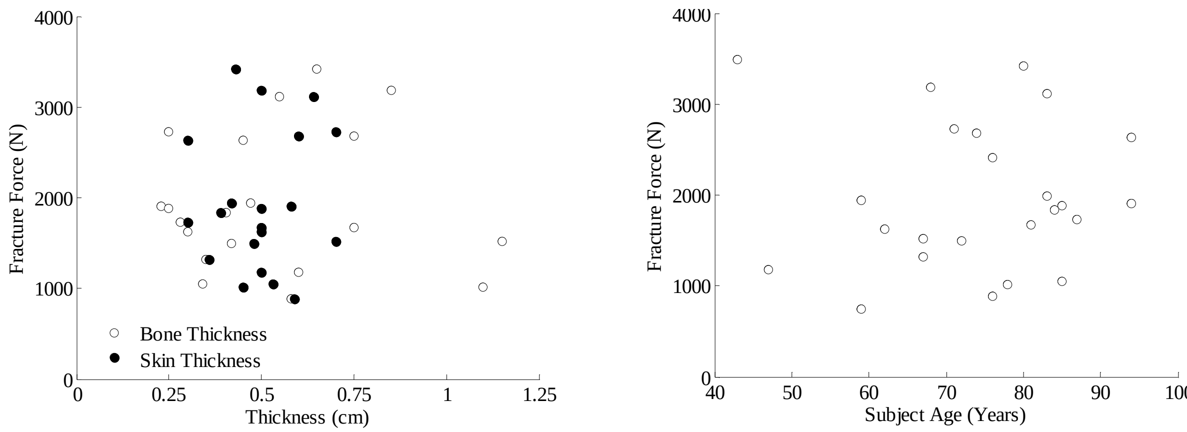

Figure 4: Frontal bone fracture force and corresponding frontal bone thickness and skin depth (left) and fracture force by subject age.

Parametric and non-parametric models were used to estimate the risk of fracture as a function of impactor force (Figure 5). Only one test per subject was used and no repeated tests resulting in fracture were used in the risk estimate. When included in the survival model, age did not statistically (p = 0.55) improve the fit. The risk does not reach 100% due to the presence of non-fracture tests at 5236, 5402, 5934 and 7613 N.

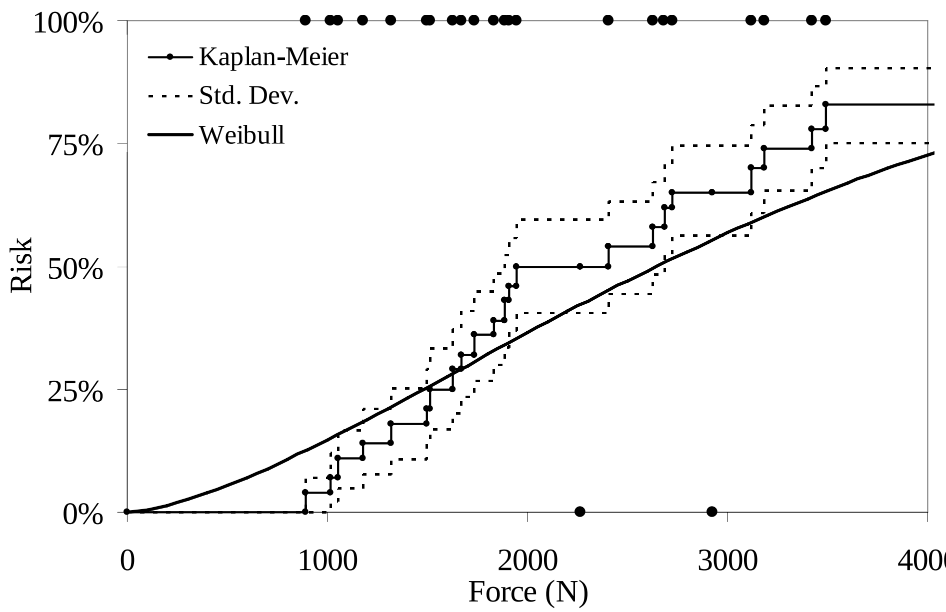

Figure 5: Risk functions for frontal bone fracture using fracture and non-fracture tests.

To demonstrate the effects of the high non-fracture tests, a separate set of risk curves were generated using only the fracture data (Figure 6). The 50% risk of fracture occurred at 1830 N for the fracture only data and 1950 N when all data were included.

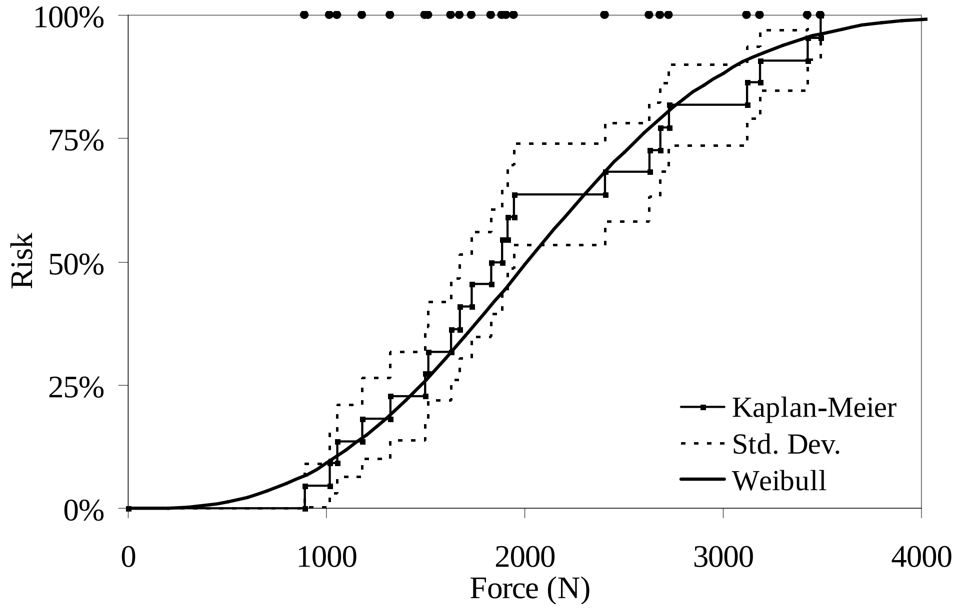

Figure 6: Risk functions for frontal bone fracture using fracture tests only.

The corridor determined from the FOCUS force-displacement response was well within the cadaveric corridors for frontal bone impacts (Figure 7). On average, the toe region for the FOCUS test exhibited a less stiff response than the cadaver response. The stiffness of the FOCUS headform increased after approximately 2 mm such that the force achieved was well within the cadaver corridor. The initial stiffness and toe region up to 20% of peak force were statistically different between the cadaver and FOCUS responses (respectively, $p = 0.001$, $p < 0.0001$). The secondary stiffness, between 20 and 80% of peak force was also statistically different ($p = 0.003$).

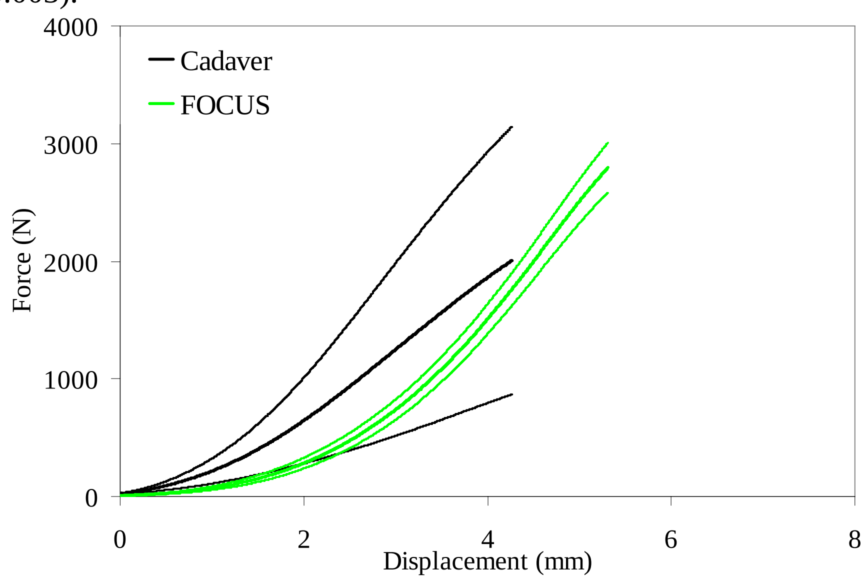

Figure 7: Corridor for cadaveric and FOCUS frontal bone response to blunt impact.

CONCLUSIONS

The purpose of this study was to provide the information necessary to understand the frontal bone response to blunt impact and determine a relationship for fracture risk. Force-deflection corridors were determined to define this relationship up to peak force. Accounting for severity, the peak force generated in each test was not influenced by the occurrence of fracture. This exemplifies the importance of using AE to determine the force at which fracture begins. The peak forces reported by Nahum *et al.* (1965, 1975) suggest that fracture actually occurred at a much lower force and, therefore, incorporation of these data in a statistical analysis should assume that they are left-censored.

The CT imaging facilitated the identification of the frontal sinus and its location with respect to the impact site. In eleven of the 22 subjects with CT imaging, the fontal sinus was present at the impact location. This presents a fundamentally different structure that consists of a relatively thinner anterior structure compared to the frontal bone. Interestingly, the frontal sinus is not symmetric about the midline of the skull and, therefore, one side of the frontal bone may exhibit a fundamentally different response than the other. In this study, the majority of subjects without a frontal sinus within the impacted region did not sustain a depressed fracture. These subjects typically incurred radiating fractures with little or no depression. A separate estimate of risk was created using the 22 subjects and it was found that the presence of the frontal sinus within the impacted region was a statistically significant parameter in the model (p=0.0422). The risk estimate generated with the reduced dataset does not incorporate all the data available, therefore it does not represent the distribution of all data collected. However, it does demonstrate the influence of the frontal sinus on fracture risk with corresponding fracture risks at 1600 N with a sinus present and 2500 N when a sinus was not present at the impact site.

A series of tests were performed to assess the response of the FOCUS headform to frontal bone impacts. The force-displacement response of the FOCUS was within the cadaver corridors defined by the characteristic average, however, there were statistically significant differences between the stiffness values of the cadaver and FOCUS headform. This study provides the data necessary to improve the biofidelity of the frontal bone region of the FOCUS headform to improve its utility in predicting facial fractures.

ACKNOWLEDGEMENTS

The authors would like to recognize the students of the Transportation Research Center of Ohio State for their outstanding research talent and thank them for their generous help.

REFERENCES

ALLISON, P. D. (1995). Survival Analysis Using SAS: A Practical Guide. Cary, SAS Press.

ALLSOP, D. and KENNETT, K. (2002). Skull and Facial Bone Trauma. Accidental Injury: Biomechanics and Prevention. A. Nahum and J. Melvin. New York, Springer-Verlag. 2nd Edition**:** 254-76.

ALLSOP, D., WARNER, C., et al. (1988). Facial Impact Response – A Comparison of the Hybrid 3 Dummy and Human Cadaver. Society of Automotive Engineers SAE No. 881719.

BERMOND, F., KALLIERIS, D., et al. (1989). Human Face Response at an Angle to the Fore-aft Vertical Plane Impact. Proceedings of the IRCOBI.

BRUYERE, K., BERMOND, F., et al. (2000). Human Maxilla Bone Response to 30 degree Oriented Impacts and Comparison with Frontal Bone Impacts. Proceedings of the 44th Association for the Advancement of Automotive Medicine Conference: 219-34.

CANTOR, A. B. (2003). Analysis Techniques for Medical Research. Cary SAS Press.

CESARI, D., RAMET, M., et al. (1989). Experimental Evaluation of Human Facial Tolerance to Injuries. Proceedings of the IRCOBI, Stockholm, Sweden.

CORMIER, J., MANOOGIAN, S., et al. (2008). The Use of Acoustic Emission in Facial Fracture Detection. Biomed. Sci. Instrum. 44: 147-52.

FUNK, J., CRANDALL, J., et al. (2002). The Axial Injury Tolerance of the Human Foot/Ankle Complex and the Effect of Achilles Tension. Journal of Biomechanical Engineering 124.

FUNK, J., SRINIVASAN, S., et al. (2002). The Effects of Axial Preload and Dorsiflexion on the Tolerance of the Ankle/Subtalar Joint to Dynamic Inversion and Eversion. Stapp Car Crash Journal 46(0S2-34).

HODGSON, V. R., LANGE, W. A., et al. (1965). Injury to the Facial Bones. Proceedings of the 9th Stapp Car Crash Conference SAE No. 650957.

KENT, R., STACEY, S., et al. (2008). Dynamic Pinch Tolerance of the Phalanges and Interphalangeal Joints. Traffic Inj Prev 9(1): 83-8.

KENT, R., STACEY, S., et al. (2008). Dynamic Pinch Tolerance of the Phalanges and Interphalangeal Joints. Traffic Injury Prevention 9: 83-88.

KLEINBAUM, D. and KLEIN, M. (2005). Survival Analysis: A Self-Learning Text. New York, Springer Science.

LESSLEY, D., CRANDALL, J., et al. (2004). A Normalization Technique for Developing Corridors from Individual Subject Responses. Society of Automotive Engineers 2004-01-0288.

NAHUM, A. M. (1975). The Biomechanics of Facial Bone Fracture. Laryngoscope 85(1): 140-56.

NAHUM, A. M., GATTS, J. D., et al. (1968). Impact Tolerance of the Skull and Face. *Proceedings of the 12th Stapp Car Crash Conference*, New York.

NYQUIST, G. W., CAVANAUGH, J. M., et al. (1986). Impact Tolerance and Response of the Face. Proceedings of the Advances in Bioengineering Conference, Anaheim, CA.

RUDD, R., CRANDALL, J., et al. (2004). Injury Tolerance and Response of the Ankle Joint in Dynamic Dorsiflexion. Proceedings of teh 48th Stapp Car Crash Conference SAE No. 2004-22-0001.

SCHNEIDER, D. C. and NAHUM, A. M. (1972). Impact Studies of Facial Bones and Skull. Proceedings of the 16th Stapp Car Crash Conference.

WELBOURNE, E. R., RAMET, M., et al. (1989). A Comparison of Human Facial Fracture Tolerance with the Performance of a Surrogate Test Device. Proceedings of the 12th ESV, Gothenberg, Sweden.

YOGANANDAN, N., FINTAR, F., et al. (1988). Steering Wheel Induced Facial Trauma. Proceedings of the 32nd Stapp Car Crash Conference.

YOGANANDAN, N., PINTAR, F. A., et al. (1993). Human facial tolerance to steering wheel impact: A biomechanical study. Journal of Safety Research 24(2): 77-85.

AUTHOR LIST

1. Joseph M Cormier
   5711 University Heights Suite 100
   San Antonio TX 78249
   210-691-0281
   jcormier@brconline.com

2. Jill Bisplinghoff
   Virginia Tech
   443 ICTAS Building
   Stanger Street MC 0194
   Blacksburg VA 24061
   540-231-3945
   bisplinj@vt.edu

3. Sarah Manoogian
   5711 University Heights Suite 100
   San Antonio TX 78249
   210-691-0281
   smanoogian@brconline.com

4. Steve Rowson
   Virginia Tech
   443 ICTAS Building
   Stanger Street MC 0194
   Blacksburg VA 24061
   540-231-3945
   srowson@vt.edu

5. Anthony Santago
   Virginia Tech
   443 ICTAS Building
   Stanger Street MC 0194
   Blacksburg VA 24061
   540-231-3945
   acsantag@vt.edu

6. Craig McNally
   Virginia Tech
   443 ICTAS Building
   Stanger Street MC 0194
   Blacksburg VA 24061
   540-231-3945
   cmcnally@vt.edu

7. Stefan Duma
Virginia Tech
443 ICTAS Building
Stanger Street MC 0194
Blacksburg VA 24061
540-231-3945
duma@vt.edu

8. John Bolte
Injury Biomechanics Research Laboratory
333 West 10th Ave
3024 Graves Hall
Columbus OH 43210