

Biomechanical Response of the Human Face and Corresponding Biofidelity of the FOCUS Headform

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ABSTRACT

In order to evaluate a human surrogate, the human and surrogate response must be defined. The purpose of this study was to evaluate the response of cadaver subjects to blunt impacts to the frontal bone, nasal bone and maxilla. Force-displacement corridors were developed based on the impact response of each region. Variation in the force-displacement response of the cadaver subjects due to the occurrence of fracture and fracture severity was demonstrated. Additionally, impacts were performed at matched locations using the Facial and Ocular Countermeasures Safety (FOCUS) headform. The FOCUS headform is capable of measuring forces imposed onto facial structures using internal load cells. Based on the tests performed in this study, the nasal region of the FOCUS headform was found to be the most sensitive to impact location. Due to a wide range in geometrical characteristics, the nasal impact response varied significantly, resulting in wide corridors for human response. The FOCUS headform was found to be repeatable and produced similar forces to the cadaver subjects for the frontal bone and maxilla impact locations.

INTRODUCTION

The four most common sources of facial trauma are Motor Vehicle Collision (MVC), assault, sports and falls [1,2,3,4,5]. Facial fractures incurred due to a MVC are more likely to result in associated head injury; therefore, the identification of parameters associated with facial fracture will reduce the likelihood of more severe head injuries [3]. A recent NASS-

CDS analysis found that in frontal impacts the most common facial injuries were abrasions (21%), contusions (32%), lacerations (32%) and fractures (11%) [6]. Cormier and Duma (2009) also found that the nasal bone and maxilla are the most frequently fractured region in frontal impacts, followed by the mandible and the frontal bone.

Due to the importance of the facial structures and exposure to injury-producing events in automotive collisions, there have been several studies focused on the tolerance and stiffness of the human face. The literature concerning facial impacts consists of two main categories. One involves the use of a cylindrical impactor to simulate facial impacts with an automotive steering wheel [7,8,9,10,11,12,13]. The second category involves the use of a flat, round impacting surface [14,15,16,17,18]. An advantage of the flat impactor is that it is usually smaller and applies an impact that is isolated to a single facial bone or region. Cylindrical impacts are typically performed by striking across the entire face. Among the previous these studies on facial tolerance, few have determined the stiffness of the face. A study by Nyquist et. al. (1986) demonstrated the force-displacement relationship of the nasal bone and sub-orbital ridge due to impact with a cylindrical object [18]. The authors found that the stiffness response was characterized by a concave-up shape which could be described by a quadratic equation. Allsop et al. (1988) used a cylindrical impactor to determine the tolerance and stiffness of the maxillary and frontal bone regions. The impactor spanned the width of the face and the impacts resulted in an average stiffness of 120 N/mm for the maxilla and 1,000 N/mm for the frontal bone regions. The same study

also included matched testing performed on a Hybrid III dummy which showed similar stiffness between their cadaver subjects and the frontal region of the Hybrid III. Additional work on creating a biofidelic headform for the Hybrid III was performed by combining the data obtained by Nyquist et. al. (1986) and Allsop et al. (1988) [19]. This study discusses the development of the GM Hybrid III Deformable Face and the results of initial testing. The force-displacement requirement for the facial structure was based on the response of the zygoma region in two previous studies [7, 18]. The authors found that additional biomechanical data were necessary to fully describe the cadaver response. The development of various load-sensing face designs for the Hybrid III has been demonstrated in previous studies [20,21,22]. The authors demonstrated promising results, but had few data to base their biomechanical assessment on. The Load Sensing Face (LSF) developed by Planath and Nilsson (1988) was similar to the Hybrid III head but lacked a biofidelic response and, therefore, its force-displacement response was not similar to that of a human. The deformable load-sensing face developed by Perl et. al. (1989) consisted of six regions of the face capable of measuring loads separately. The response of the face was compared to similar impacts performed by Allsop (1988). The authors observed that the surrogate face did not exhibit the linear stiffness necessary to model the cadaver response. In all the previous work developing a biofidelic facial surrogate, the lack of cadaveric data was a limiting factor.

Evaluating safety measures to reduce the risk of facial fractures in automotive and military applications requires the use of a biofidelic surrogate. One such surrogate is the Facial and Ocular Countermeasures Safety (FOCUS) headform. The FOCUS headform has the ability to measure forces applied to the facial and orbital structures. It has been developed by Virginia Tech and Denton ATD and validated for eye injury prediction [23]. The headform is comprised of eight separate sensing regions within the facial structures that overlie embedded load cells (Figure 1). These regions correspond to the frontal bone, nasal bone, zygoma, maxilla and mandible. Each “bone” or reaction surface is made of a rigid urethane plastic that is attached to a load cell which is supported by an internal aluminum frame. The outer skin, eyes and nose insert are made of a soft urethane. Each load cell is capable of measuring forces in three axes.

<figure 1 here>

The goal of this study is to develop an understanding of the cadaveric response to blunt impact and compare this response to the FOCUS headform. Defining force-deflection corridors for facial impacts in cadavers will provide a basis for the implementation and biofidelic evaluation of the FOCUS headform or other facial surrogates.

METHODS

The data for this study were obtained by performing facial impacts on male cadaveric subjects and the FOCUS headform. The methods of this study consist of striking the facial bones with the flat face of an unpadded, free-falling cylindrical impactor. All heads were frozen and thawed prior to testing. A total of 31 male subjects ranging in age from 43 to 94 years were included in the study. Prior to testing, 15 of the subjects underwent CT imaging which was used to measure the skin thickness underlying the impacted regions.

SPECIMEN PREPARATION

The specimens were removed from the body and prepared by removing the scalp overlying the occipital region. Metal screws were inserted into the occiput to provide additional structure for the casting material to adhere to. Each head was then rigidly mounted to a semi-circular, polycarbonate support using Bondo (Figure 2). The influence of the mounting procedure on the stiffness of the skull was minimized by limiting the lateral support provided to the posterior aspect of the skull. Consistent orientation between subjects was obtained by vertically aligning the Frankfort plane prior to mounting.

CADAVER TEST CONDITIONS

Each impact was performed using a cylindrical, free-falling rigid aluminum impactor (3.2 kg) with a steel tip (Figure 2). The flat impacting surface had an area of 6.45 cm² (1 in²) and was machined with a slight bevel to reduce edge effects. Specific anatomical landmarks were used to determine impact location (Figure 3). For frontal bone impacts, the impactor was positioned such that its lower edge was centered to the orbit and superior to the superciliary arch. For nasal bone impacts, the impactor was centered over the palpated inferior surface of the nasal bone. For maxilla impacts the impactor was aligned vertically below the center of the orbit at a level such that the top of the impactor was just below the inferior orbital rim. This focused the impact at the maxillary sinus. In the majority of subjects, only a nasal or maxilla impact was performed in addition to a frontal bone impact. In some cases nasal and maxilla tests were performed on the same subject if it was determined that if fracture occurred, it did not influence the behavior of the remaining structures. Dissections were performed after testing to evaluate fracture patterns to ensure fracture in one region did not influence the behavior of another. Low-level frontal bone impacts were repeated on one side of two subjects. The high-energy fracture producing impact was performed on the contralateral side. In the event that fracture was not readily apparent, fracture status was determined during detailed dissections performed on each subject.

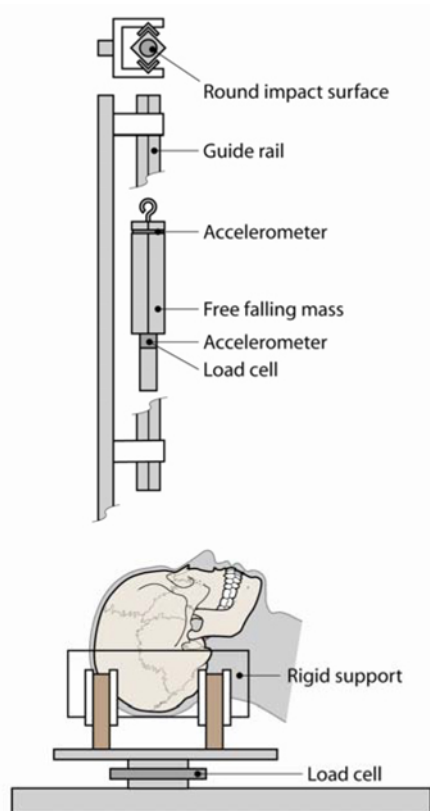


Figure 2. Schematic of test apparatus used in the current study.

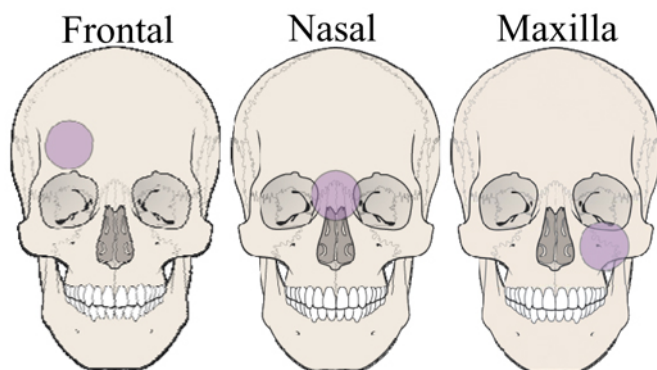


Figure 3. Impact locations used in the current study.

INSTRUMENTATION

The rigid impactor was instrumented with two single-axis accelerometers (Endevco 7264B-2000, Endevco Corp., San Juan Capistrano CA). A load cell (Denton, 8617JTF, Rochester Hills, MI) was attached to the tip of the impactor which was also instrumented with a single axis accelerometer (Endevco 7264B-2000, Endevco Corp., San Juan Capistrano CA) (Figure 2). A load cell (Denton 1968, Rochester Hills, MI) was mounted to the head support to measure reaction forces. Impact force was obtained using the impactor load

cell along with the inertially compensated tip mass. The secondary accelerometer mounted at the top of the impactor was a redundant sensor to help ensure data were obtained for each test and to compare to other sensors. Data obtained using the load cell and accelerometers were well correlated. All data were filtered to CFC 300. Previous studies have utilized CFC 180; however the use of CFC 300 did not significantly alter the measured peak forces and was chosen to increase the likelihood of capturing small changes in impactor force during fracture [8, 9, 18]. Impactor displacement was calculated by double-integrating the acceleration data. Contact between the impactor and subject was defined based on an impactor force of 10 N. Once the impactor force reached a level above 10 N, the displacement with respect to the face was set to zero and further motion was calculated by double integration. Additionally, high-speed video was also recorded at a frame rate of approximately 4,000 fps. Impactor displacement calculated by double integrating impactor acceleration was verified using high-speed video analysis. This was performed by creating a second-order polynomial fit of the displacement obtained by video analysis.

To define the average force-displacement response of each bone, a corridor was created to encompass the overall force-displacement response. The response corridor was determined by the mean and standard deviation of the characteristic average [24]. The characteristic average was created using the force-displacement response up to 90% of peak force. In cases in which there were multiple peaks, the response up to 90% of the first peak was used to generate the characteristic average. In some cases, 90% of peak force did not remove the variation in stiffness. In those cases additional data were removed to minimize the presence of a non-linear relationship. This ensured that only the linear portion of the response was included in the characteristic average calculation. The values for the corridor were determined using a custom Matlab program that implemented the algorithm described by Lessley et al. (2004). The corridor was fit using 2nd order polynomials so that the corridors could be easily recreated. Two stiffness values were calculated, at 20% (initial stiffness) and between 20% and 80% (secondary stiffness) of the peak force for each test. The toe region was defined as the displacement at which 20% of the peak force was reached. Student's t-tests were performed to test for statistical differences between the stiffness of the cadaver and FOCUS response.

FOCUS HEADFORM

A series of impacts were performed on corresponding regions of the FOCUS headform in order to characterize its response to blunt impact. Impact severity was chosen to facilitate comparison with cadaver results and to minimize risk of damaging the headform. The impact was applied using the same apparatus and instrumentation scheme used for the

cadaver impacts. The internal FOCUS load cell data were filtered to CFC 300. Forces measured by the FOCUS internal load cells were recorded and compared to impactor forces and examined for load-sharing between facial regions. Only forces measured in the anterior-posterior (x-axis) were analyzed due to the direction of the impact applied to the headform. The stiffness of the FOCUS response was determined up to 20% and between 20% and 80% of the peak force for comparison to the cadaveric response. This separation was chosen to provide a more realistic description of the force-displacement response, which consisted of a softer toe region.

RESULTS

FRONTAL BONE

Cadaver

Impactor displacement determined from double integration of accelerometer data was almost identical to that determined from high speed video (mean $r^2 = 0.98$). Due to the nature of the interaction between the frontal bone and the impactor, in some tests the impactor slid along the surface of the bone. Therefore, in these cases the force-displacement response did not reflect the deflection of the bone and soft tissue. These cases were identifiable by a low initial stiffness and excessive displacement. The lack of any rebound and continued displacement after impact was a clear indicator of impactor slip. The tests identified as exhibiting a sliding motion prior to peak force were not included in the development of a force-displacement corridor. If slippage was questionable and video analysis was not available to verify, the test was not utilized for the corridor development. On average, the thickness of the skin underlying the impacted region was 5 mm thick (std. dev.: 1 mm, min: 3 mm, max: 7 mm). There was no statistically significant correlation between skin thickness and the stiffness of the frontal bone. The dataset used to assess the force-displacement response of the cadaveric frontal bone consisted of 20 tests (Appendix A). The stiffness of the frontal bone was calculated at 20% of peak force and between 20% and 80% of peak force (Table 1).

Table 1. Stiffness of frontal bone response at 20% and between 20 to 80% of peak force.

	Stiffness (N/mm)		Displacement at 20% peak force (mm)
	20%	20-80%	
Mean	298	978	1.7
Std. Dev.	74	523	0.4
Min	186	159	1.1
Max	441	2200	2.6

The force-displacement response varied depending upon the type of fracture incurred by the specimen. Tests resulting in a

depressed fracture tended to have a force-displacement response that reached an initial peak followed by continued displacement with an oscillating force (Figure 4). In some cases the impactor continued to translate, while others stopped within the depression in the frontal bone. Tests that caused a stellate fracture pattern with little depression demonstrated a force-displacement response that reached a peak and then decreased with little continued impactor motion (Figure 5). These phenomena occurred in subjects exposed to the same impactor energy.

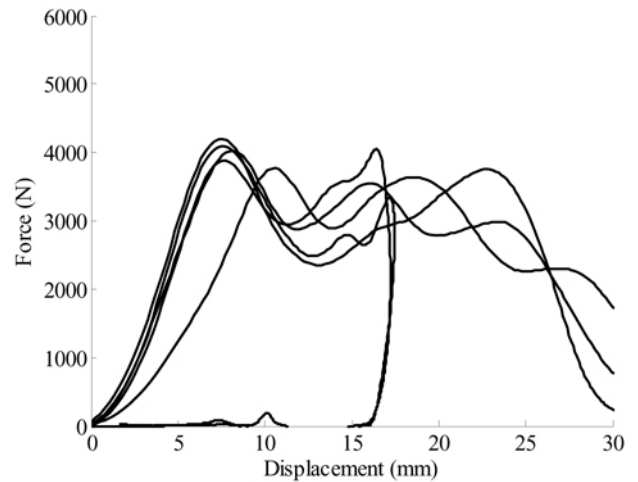


Figure 4. Representative frontal bone force-displacement responses for tests resulting in a depressed frontal bone fracture.

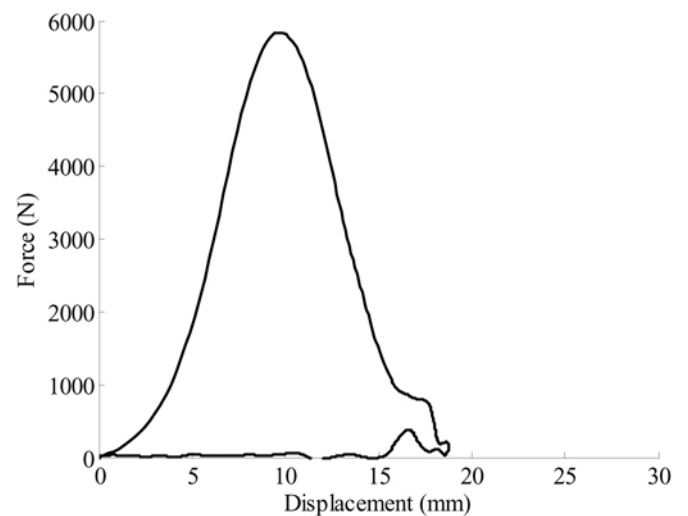


Figure 5. Frontal bone force-displacement response of test resulting in stellate fracture with no depression.

The range of the force-displacement response up to 90% of the peak force was described by calculating the characteristic average (Figure 6) [24]. The corridor was only defined for tests in which the impactor did not slide against the frontal

bone. The corridor was created by only using the standard deviation of the characteristic average in the y-axis, or force direction. This was done because it generated a tighter fit to the current data, creating a smaller corridor. The majority of tests were within one standard deviation of the mean with some tests exhibiting a stiffer response. The corridors can be accurately recreated using 2nd order polynomials (Figure 7).

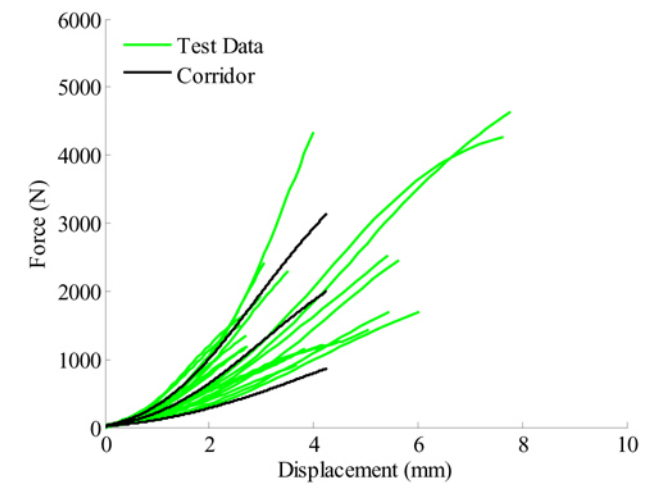


Figure 6. Frontal bone force-displacement relationship of cadaver specimens with corridor.

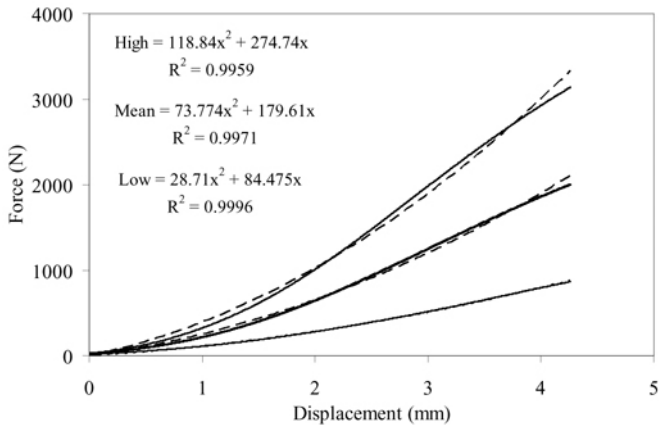


Figure 7. Polynomial fits of frontal bone response corridors.

FOCUS Headform

A total of 6 tests were performed on the frontal bone region of the FOCUS headform. Impactor velocity ranged from 2 to 2.2 m/s corresponding to an impactor energy of 6.3 to 7.9 J. The peak forces measured by the impactor ranged from 2754 to 3260 N. The peak force measured by the internal FOCUS load cells were practically equal (Figure 8) to the impactor force with an average difference of 0.3%. On average 20% of the response occurred at a displacement of 2.7 mm (Std. Dev.

= 0.18) with an initial stiffness of 210 N/mm (SD = 19) and a secondary stiffness of 846 N/mm (SD = 68).

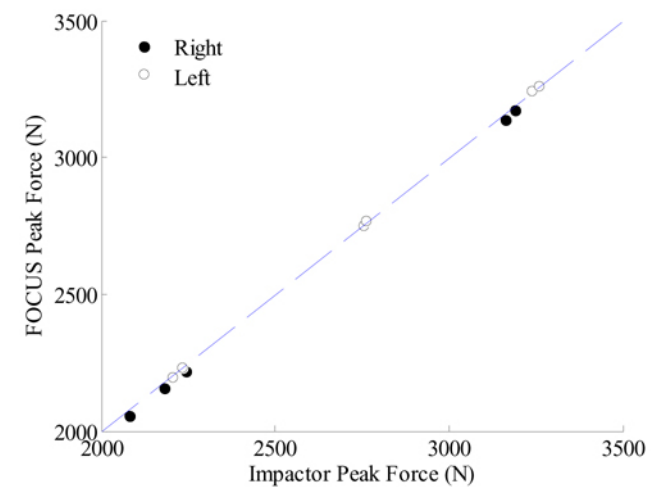


Figure 8. Impactor force and internal FOCUS force due to frontal bone impact.

The force-displacement of the FOCUS headform was repeatable and followed the same upward path at the various impactor energies. The FOCUS response to the impact was elastic and resulted in the impactor bouncing off the headform following the initial impact. The stiffness of the FOCUS response increased with impact velocity over the ranged utilized (Table 2).

Table 2. Stiffness of FOCUS frontal bone response by range in impactor velocity.

Impactor Velocity (m/s)	Initial Stiffness (N/mm)		Secondary Stiffness (N/mm)	
	Mean	Std. Dev.	Mean	Std. Dev.
2	197	11	761	2
2.2	217	20	889	22

The characteristic average of the FOCUS force-deflection response was created using all tests (Figure 9) [24]. There was no evidence within high-speed video or impactor displacement data to suggest that the impactor slid along the FOCUS headform during any test. The 2nd order polynomials fit the derived corridor well. Therefore, they can be useful in recreating the average response of the FOCUS headform (Figure 10).

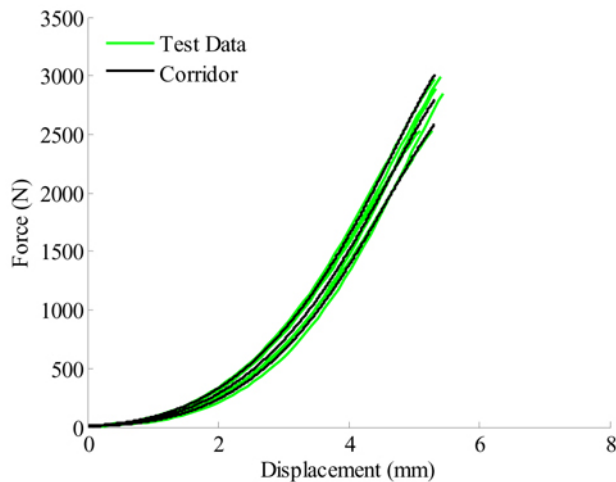


Figure 9. FOCUS headform characteristic average for frontal force-displacement.

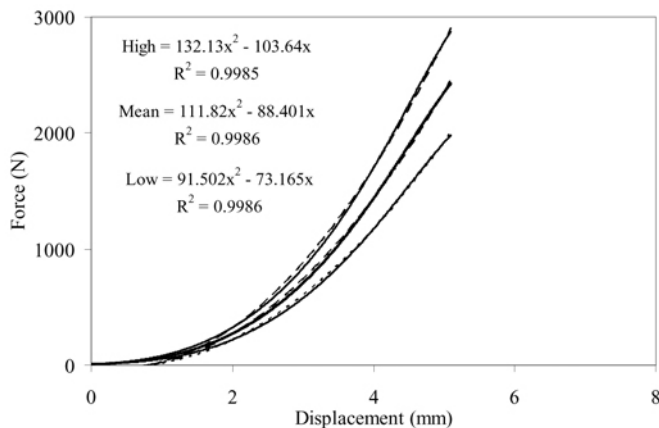


Figure 10. Polynomial fit to FOCUS frontal bone force-displacement corridor.

The corridor determined from the FOCUS force-displacement response was well within the cadaveric corridors for frontal bone impacts (Figure 11). 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 ($p = 0.003$).

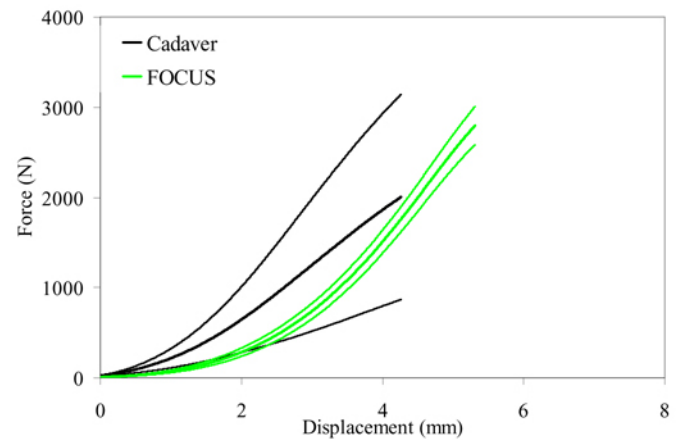


Figure 11. Corridor for cadaveric and FOCUS frontal bone response to blunt impact.

NASAL BONE

Cadaver

In all cases force-deflection relationships were determined using double integration of impactor acceleration data. Impactor displacement determined from double integration of accelerometer data was almost identical to that determined from high speed video (mean $r^2 = 0.98$). As a result of the wide variation in subject anthropometry, the displacement at which engagement of the actual nasal bone differed greatly between subjects. A total of 19 tests were used to generate the force-displacement corridor for the nasal bone (Appendix A). This created a wide variation in the displacement achieved during the toe region of the force-displacement response (Table 3).

Table 3. Nasal bone stiffness determined at 20% and between 20 and 80% of peak force.

	Stiffness (N/mm)		Displacement at 20% peak force (mm)
	20%	20-80%	
Mean	39	261	5.1
Std. Dev.	41	217	3.2
Min	8	100	0.7
Max	132	986	12.1

The majority of the tests exhibited an initial low stiffness for approximately the first 5 mm of displacement (Figure 12). Within this group, all tests resulting in fracture exhibited some degree of variation in stiffness with increasing displacement. In the majority, an initial peak in force was followed by a stiffer response to a higher force before coming to a stop. The five tests that did not fit the category above demonstrated a relative lack of a toe region and reached higher peak forces (Figure 13). All but one of these tests exhibited an initial peak in force with a subsequent higher peak at the end of impactor travel. The single test which did