AM 65-20

TOLERANCES OF THE HUMAN FACE TO CRASH IMPACT

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Federal Air Surgeon

July 1965

FEDERAL AVIATION AGENCY
Office of Aviation Medicine
Civil Aeromedical Research Institute
Oklahoma City, Oklahoma

ACKNOWLEDGMENTS

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TOLERANCES OF THE HUMAN FACE TO CRASH IMPACT

I. Introduction

Many accident investigators have reported that 70% to 80% of all deaths and injuries in crash decelerations are from face and/or head injuries caused by body flailing and the head striking surrounding structures with less yield characteristics than those of the head.[1, 3–7] Design engineers can eliminate a large portion of these injuries and deaths by structural redesign if facial and head tolerances to impact are made available to them.

Most researchers in the past have reported on studies designed to determine tolerances of the human head to impact forces applied to the top, side, and back of the head against nonyielding flat steel blocks, flat concrete, steel balls, or hammer blows. In crashes of aircraft as well as in other transportation vehicles, the head usually contacts structures that have some degree of deformation yielding. This bending increases the deceleration time history and allows the structure to contact more surface area of the face or forehead, both of which should increase the tolerance limits of the head. Snyder's study of survival in fall cases,[8] studies of Gurdjian et al., of cadaver impacts against laminated glass,[2] and the early work done at Cornell shooting plastic skulls at deformable panels[9] (all against yielding structures) seem to support this theory, in a general way. The purpose of the work presented here is to delineate tolerances of each portion of the face and forehead to serve as guidelines for engineers in the design of structures in our transportation environments that would produce less injury upon impact.

II. Procedure and Discussion

To obtain impact tolerance data of living human heads, auto salvage yards were searched to locate dash panels that had dents made by the head of the right front seat passenger as he jacknifed forward during a front-end collision. These dash panels were purchased from the salvage yard and brought to the laboratory for study. Since the auto license plate remains with the car in Oklahoma, it was possible to contact the injured passenger and obtain his complete medical history.

Next, entire cowlings from identical cars (make, model and year) were purchased and impacted with an instrumented dummy head on a small catapult (Figure 1. All figures, 1 through 20, are in the Appendix), until the dent made by the human head was duplicated as to area, depth, and exact location on the dash. Thus, injuries sustained in producing the head dent could be correlated with the "g" time—force parameters from the dummy-head impacts to determine forces necessary to produce unconsciousness, facial fractures, and lacerations.

Certain facts became apparent early in this study. The facial structure is, roughly, a convex portion of a sphere, whereas the automobile dash panel is a portion of a cylinder lying on its side and convex toward the passenger. The initial contact point of these two surfaces during impact is necessarily small in area. If the dash panel is weaker than the portion of the face making contact, it deforms, contours to the face, and increases the contact area and distribution of the impact load. Thus, in a number of cases studied, there was a progressive failure of facial structure, and it was impossible to separate the fracture forces for separate facial bones. More was learned from the nonfracture cases, and, although more than one facial bone made contact, it was possible to tell accurately from lacerations and bruises the area of the face taking the impact load. Some of the most interesting cases are presented in Figures 2 through 9, and the total data are summarized in Figures 19 and 20.

To verify the data obtained in the accident study and to determine the fracture points of each facial structure separately, 45 impacts were made on cadaver heads using the small catapult. Blocks containing accelerometers (to record input forces) were *molded to fit* the individual facial bones and placed against the face. The face was then impacted with increments of force until fracture occurred. The areas studied in this manner along with fracture tolerances are shown in Figures 19 and 20. A second accelerometer was rigidly attached to the back of the skull (using dental acrylic) to record the magnitude and duration of the input loads transmitted through the head. Figures 10 through 16 show typical fractures with flesh removed and the shapes of input and output loads.

In order to determine whether the forces required to fracture individual bones of the face would be additive if the forces were applied to the entire face simultaneously, a final test was made. In this test, a cadaver head was impacted against a block molded to fit the entire surface of the face (Figure 17). A force of over 300 g, the highest attainable with our catapult, caused no lacerations or injuries. Response during impact of facial tissue of cadaver heads would necessarily be different than for living tissue. The heads used in this study, however, were in an excellent state of preservation with facial tissues pliable and moist, and the author feels the effects of tissue differences on facial tolerances were negligible.

The cadaver heads tested in this study were from persons whose age ranged from 28 to 74 years of age with a mean age of 51. Before testing, the cranial cavity left by shrinkage of brain tissue was filled with a gelatin material (same specific gravity as brain) through the *foramen magnum.* After adding the gelatin, head weights varied from 8.50 to 11.82 pounds, with a mean of 10.39 pounds.

Although individuals vary in tolerances to fracture forces, there was no correlation between tolerance and age. The frontal bone of one 66-year-old head fractured with a force of 330 g on the 2½-sq. in. block, while the frontal bone of a younger (38-year-old) head fractured at 190 g on the same area block.

III. Conclusions.

There is a shameful and needless loss of life and facial destruction in crash impacts with transportation vehicles. Man, in a vehicle, is surrounded by rigid tubes, angles, knobs, heavy door posts, sharp instruments, and heavy metal of small radius of curvature (to name a few) all designed to impact the face and head on very small areas.

This study has shown that if this environment were changed to a medium-weight deformable metal (without heavy structure directly behind it) with a radius of curvature of 6 to 10 inches for energy attenuation and padded with 1 to 2 inches of slow return material to contour to the bones of the face and distribute the impact load over the available area of the face, it would be impossible to produce facial and forehead fractures in crash impacts. The limit of human tolerance would then be the forces necessary to produce brain lacerations without fracture.

As might be suspected, the weakest part of the face is the nose, which has a fracture point varying between 35 and 80 g. Impact force on a single zygomatic prominence of 50 to 80 g will produce compound fractures of the arches. Condyles of the mandible will be fractured by forces of between 70 and 110 g applied on the tip of the chin. The teeth and maxilla can withstand forces of more than 150 g if applied to a contoured area of about 4 sq. in. The anterior surface of the cranium (forehead) is more rugged, requiring forces ranging between 120 and 180 g (applied to 1 sq. in.) to produce fracture. Utilizing 3 to 4 inches of the forehead area as the impact point raised the tolerance to as high as 300 g in some tests. Automobile accident injuries studied here established that blows to the face in excess of 30 g produce unconsciousness (15 minutes to 2 hours) with or without fractures. Variations were noted between the fracture tolerances of different heads, but these variations did not correlate with age.

Airline-seat and aircraft manufacturers should design all structures surrounding the passengers to deform with head impacts of 40 ft/sec and not exceed this 30-g figure. In addition, satisfactory padding should be provided to distribute the impact load over as much facial area as possible.

Designers of other transportation vehicles should strive not to exceed 40 g since temporary unconsciousness is not such a major concern in escape.

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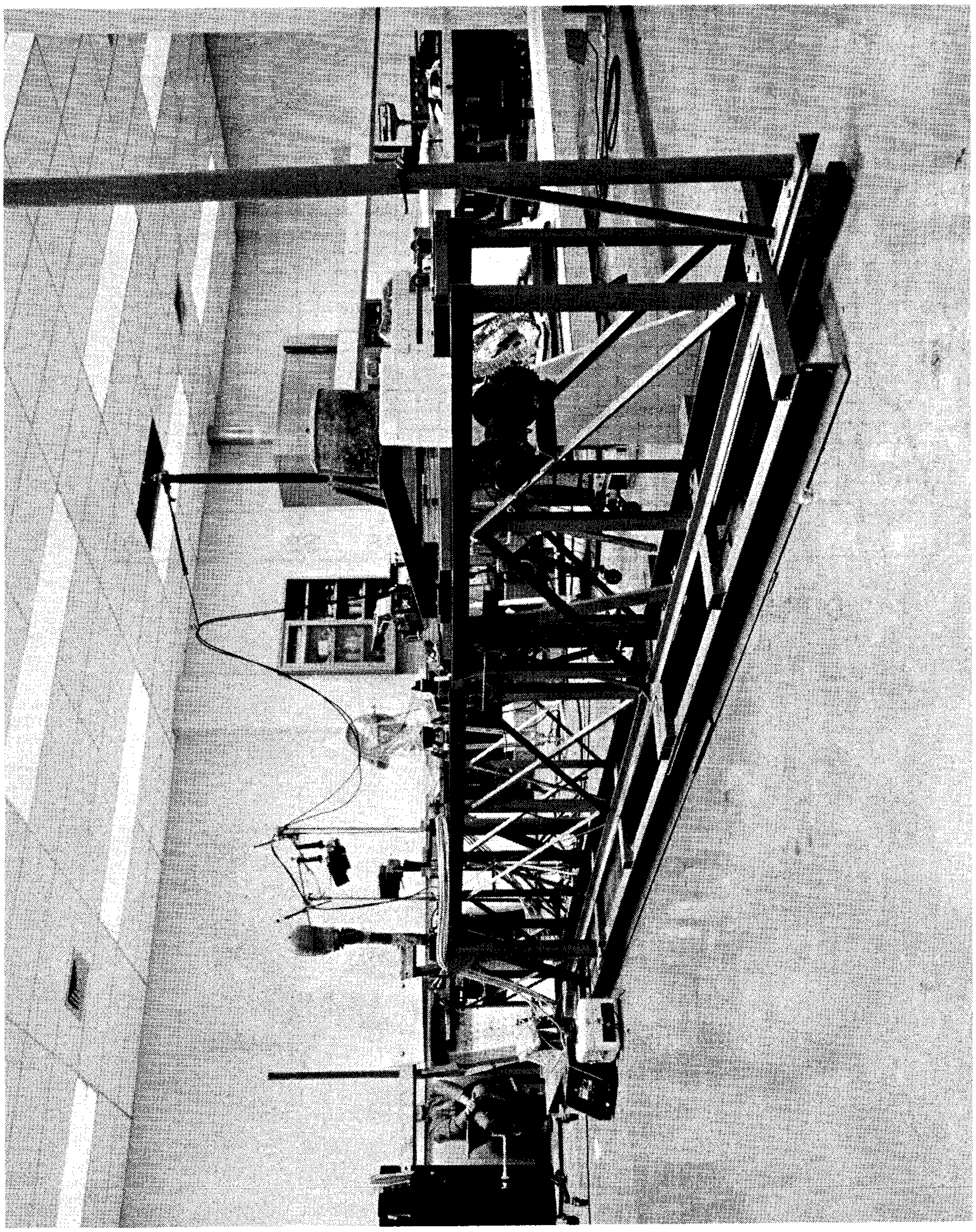


FIGURE 1. Overall view of head-impact catapult. Rubber bungee cords are stretched taut to accelerate sled up to speeds of 100 m/hr. Sled is stopped at desired point by fricton brake allowing head and upright arm to swing forward freely on ball-bearing rollers. The forward arc simulates body motion produced by the snubbing action of a seat belt.

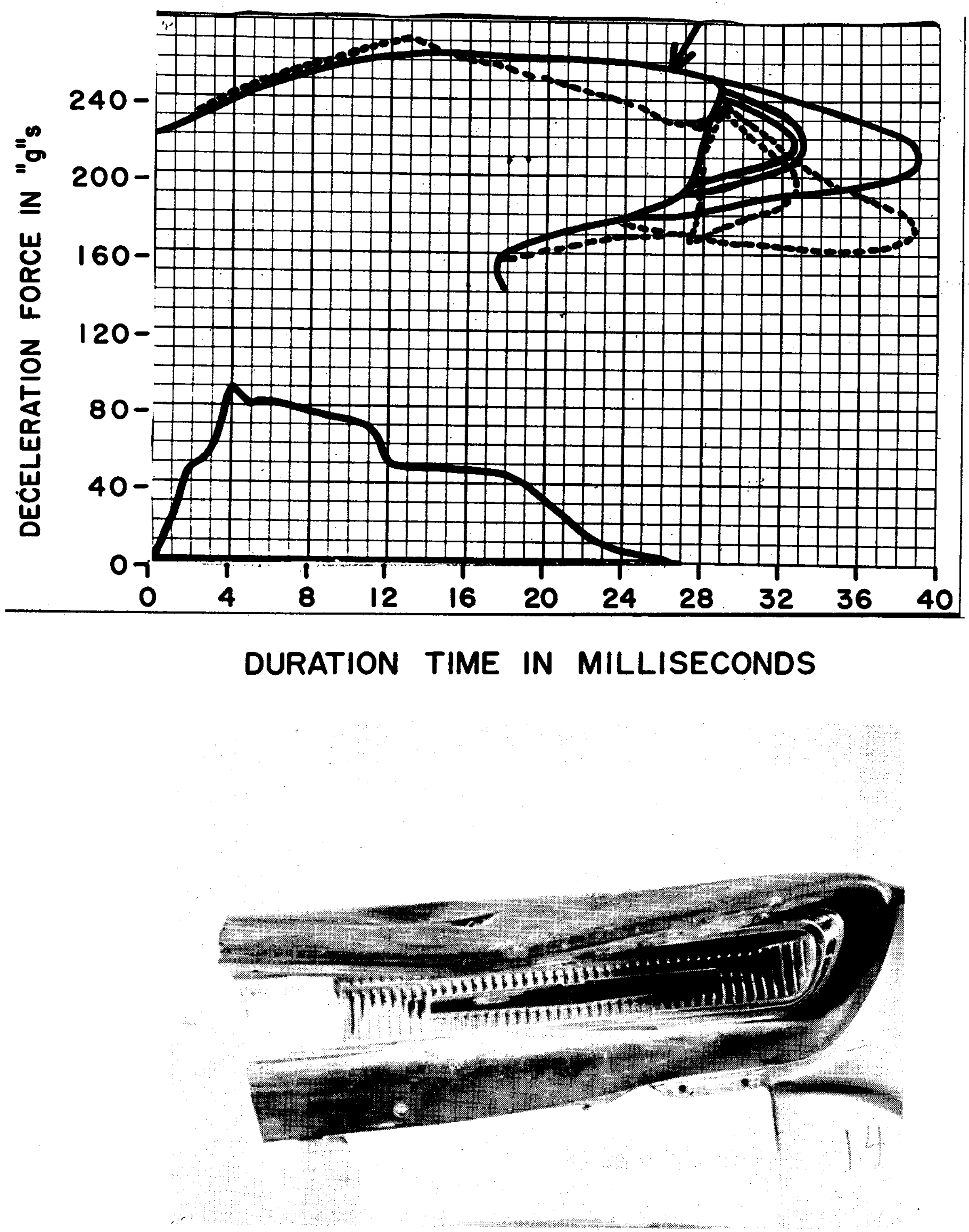
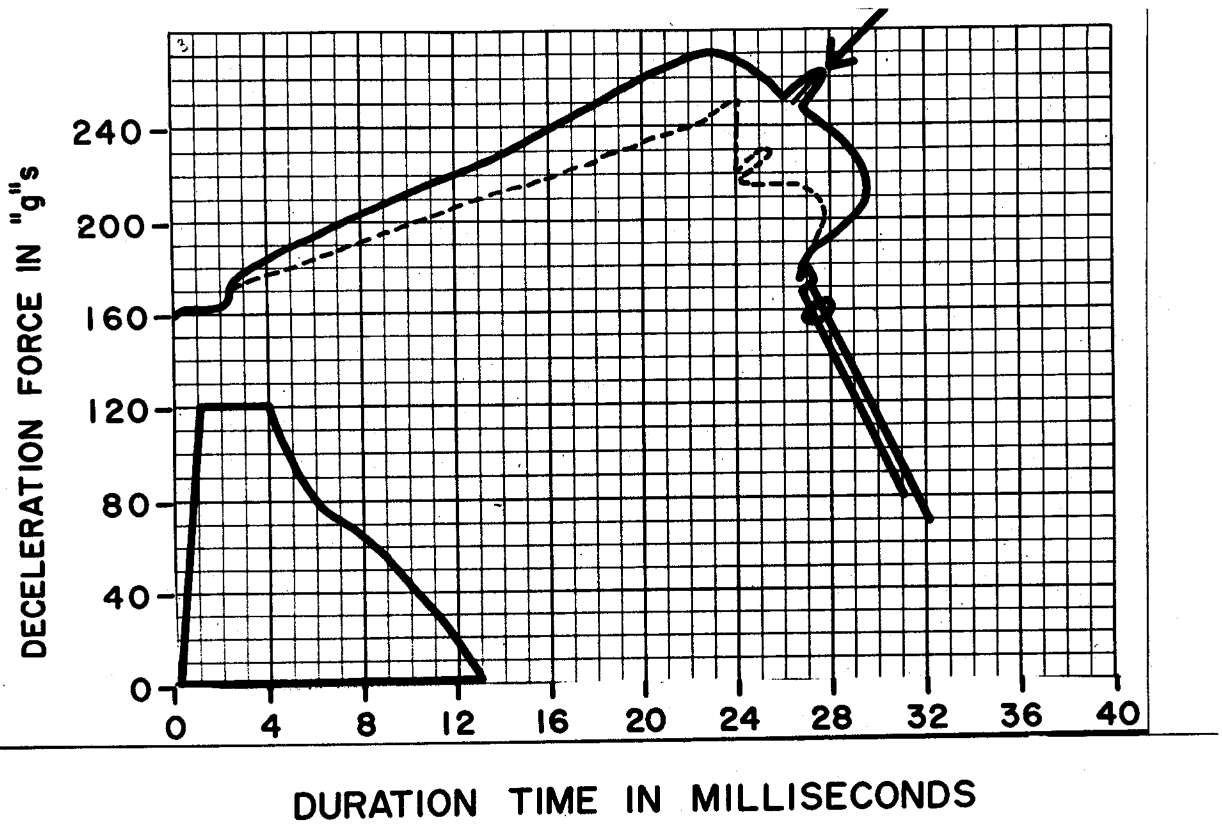


Figure 2. Forehead impact producing lacerations and concussion.



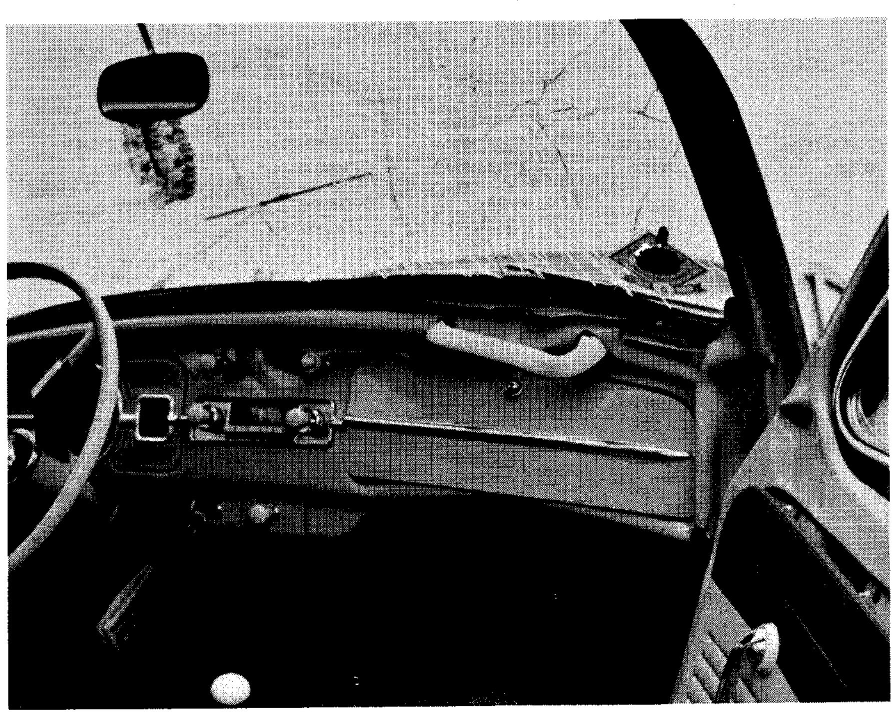
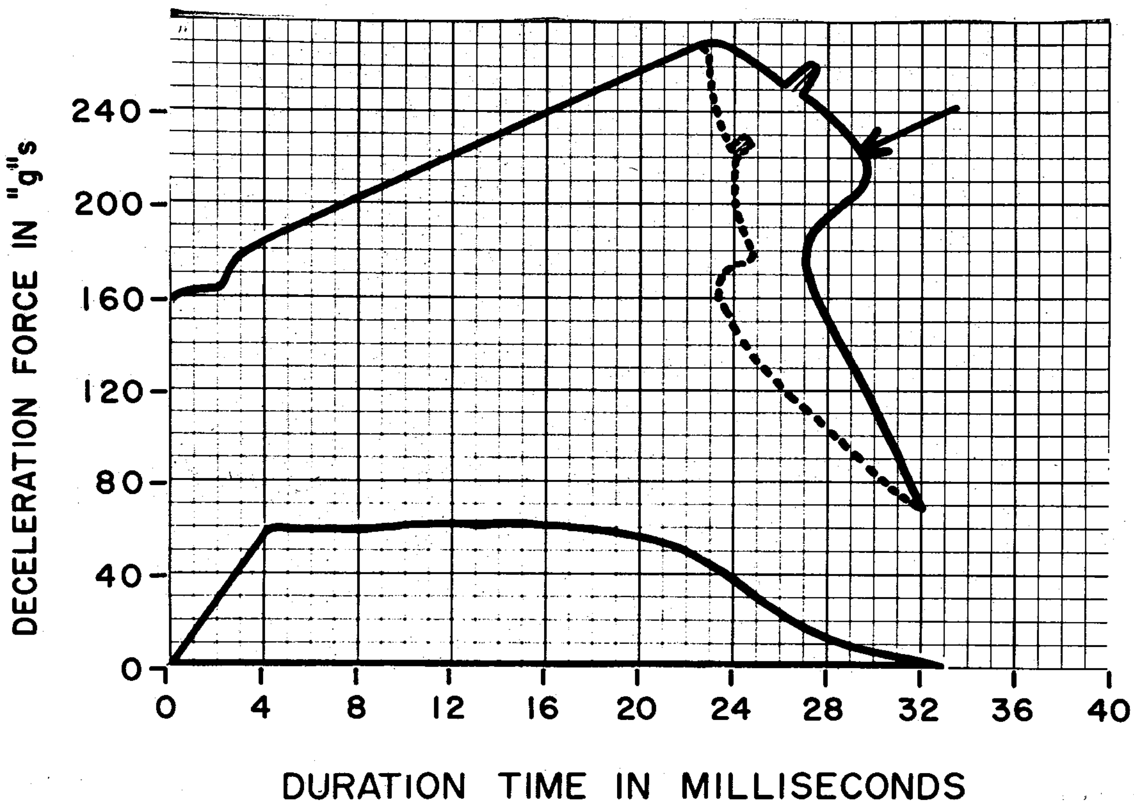


FIGURE 3. Forehead impact against a sharp rigid edge. Fatal skull fracture of frontan bone.



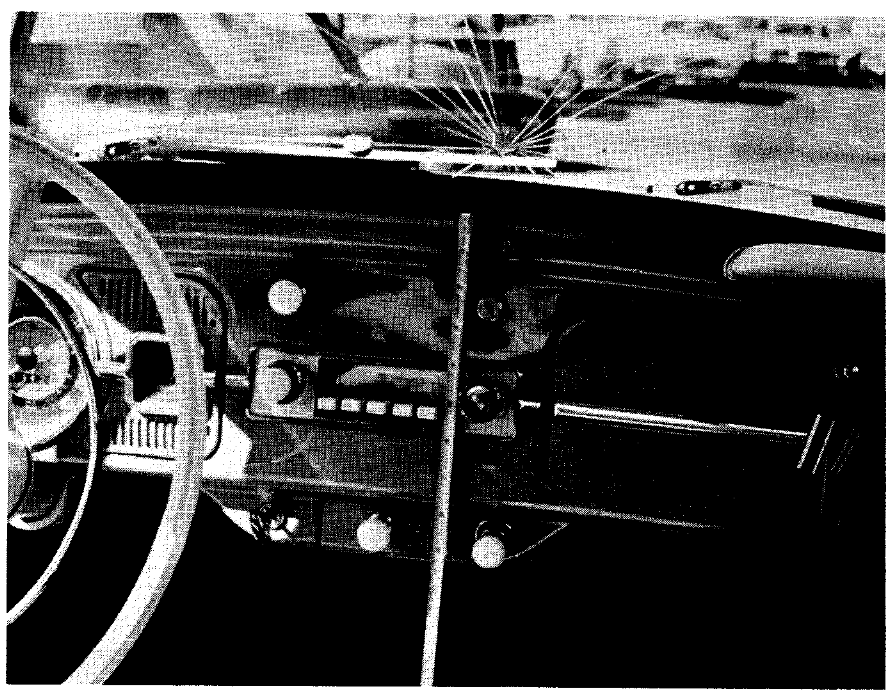
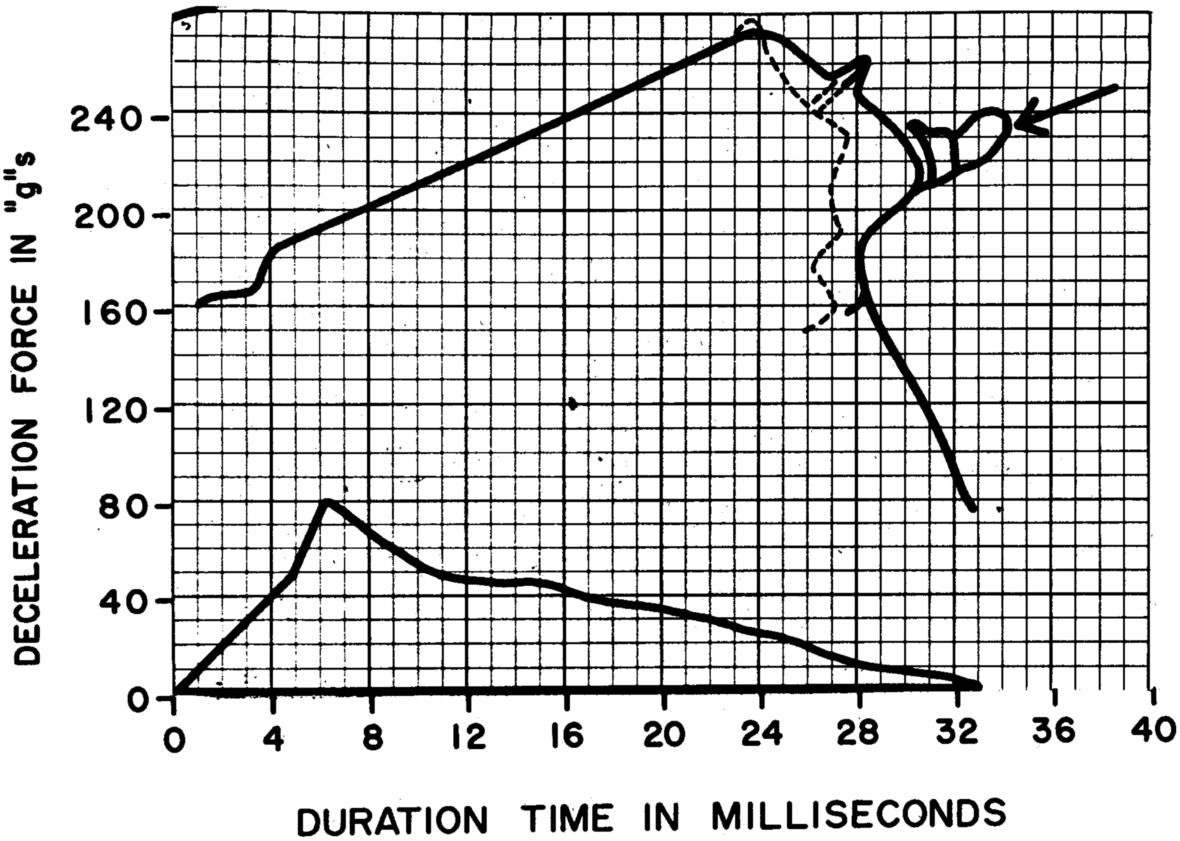


FIGURE 4. Nose and mouth impact of a 7-year-old girl. Laceration of mouth and nose, no fractures.



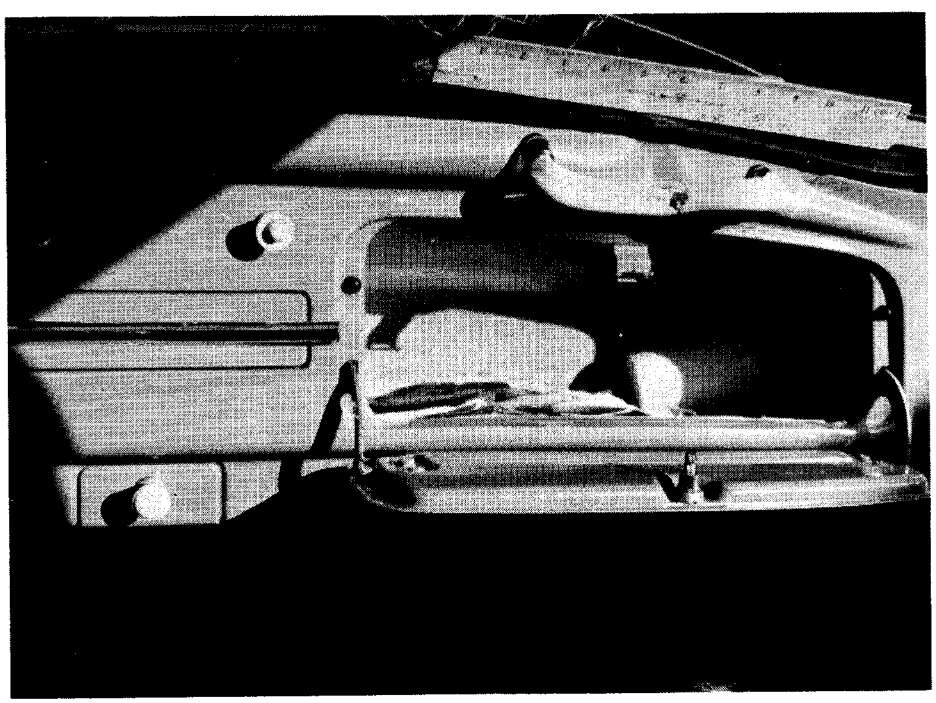
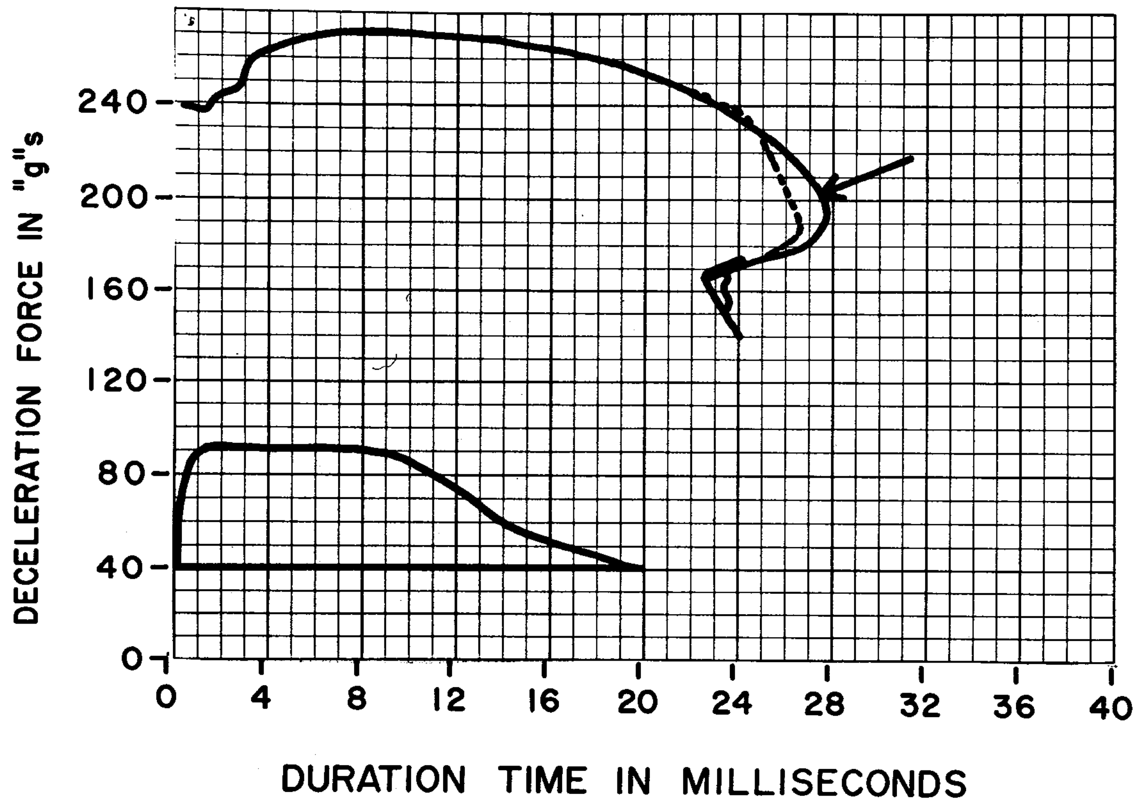


FIGURE 5. Nose impact. Severe lacerations and fracture.



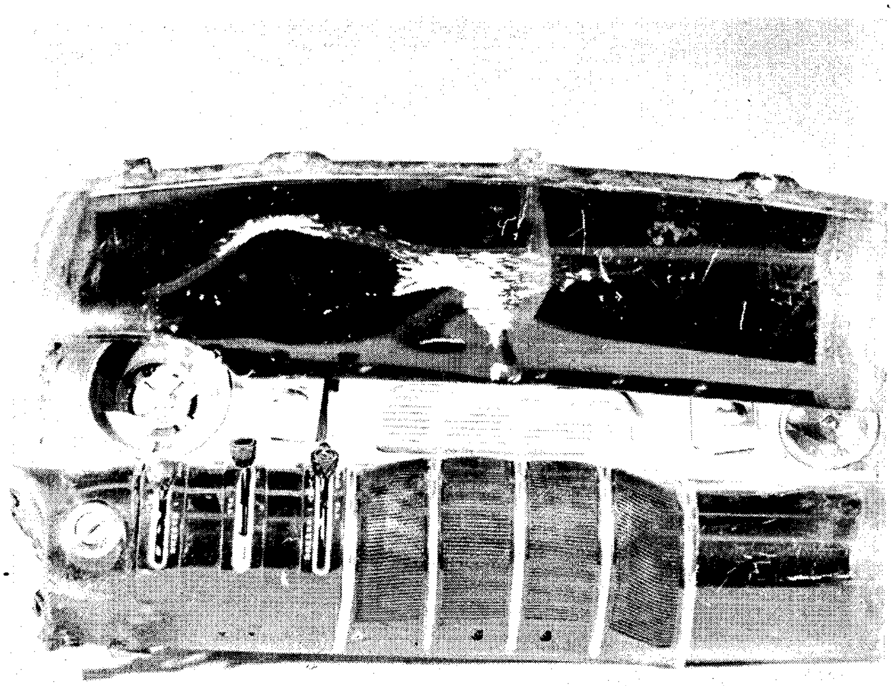
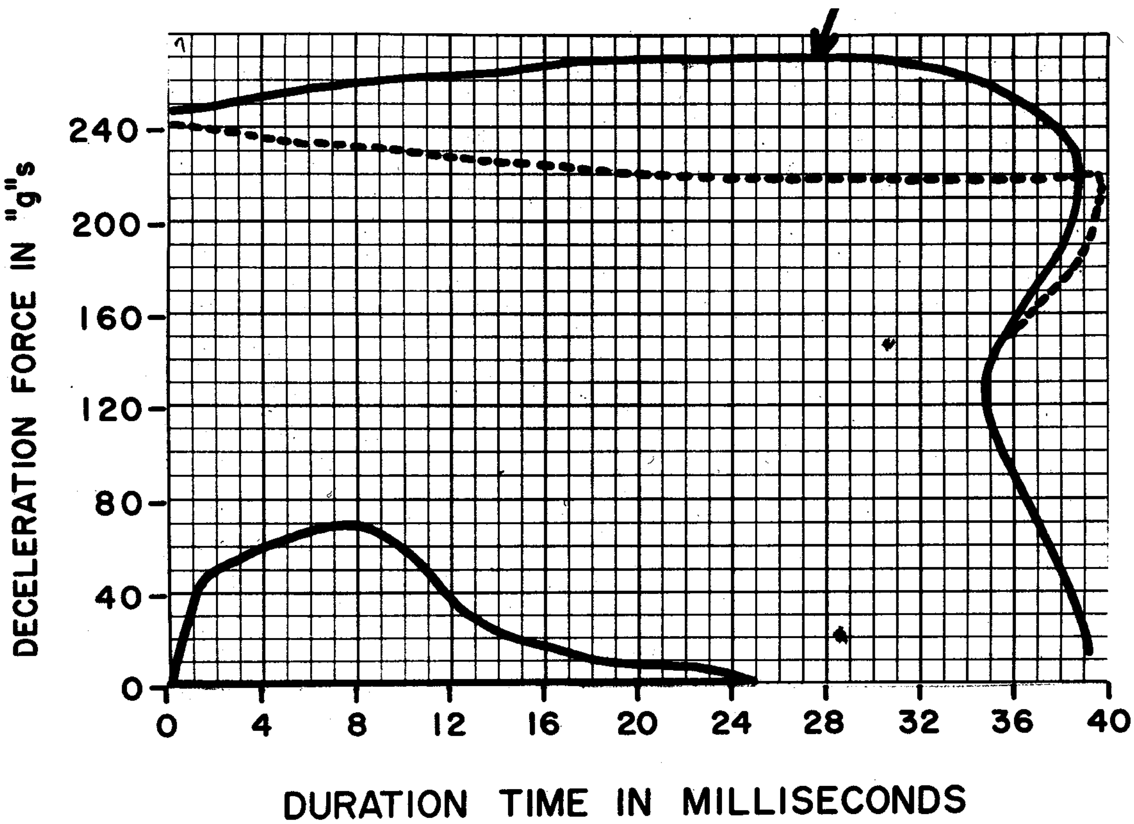


FIGURE 6. Impact of right zygomatic arch. Bruises and concussion.



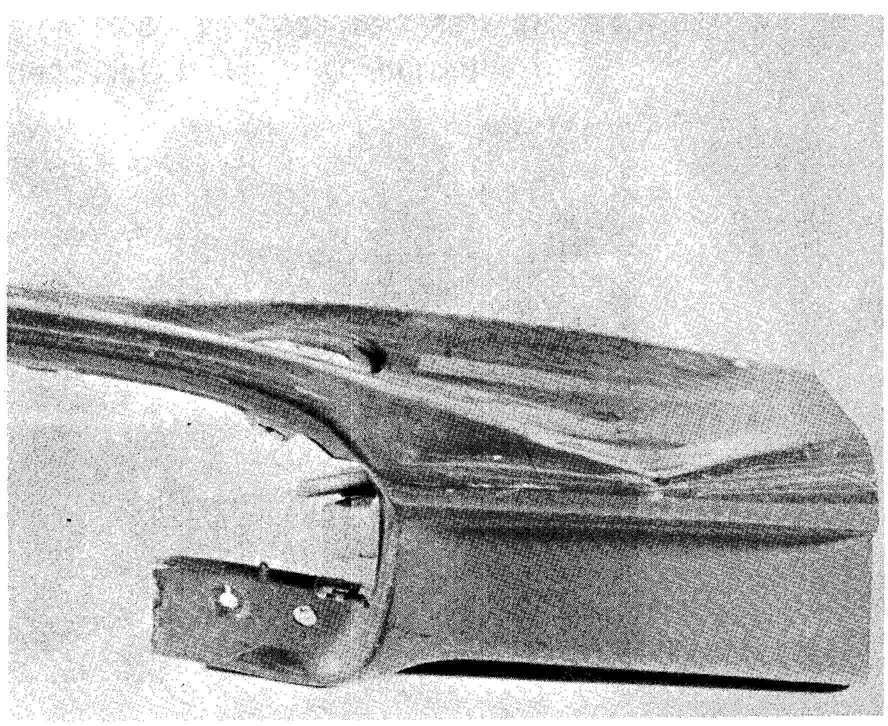
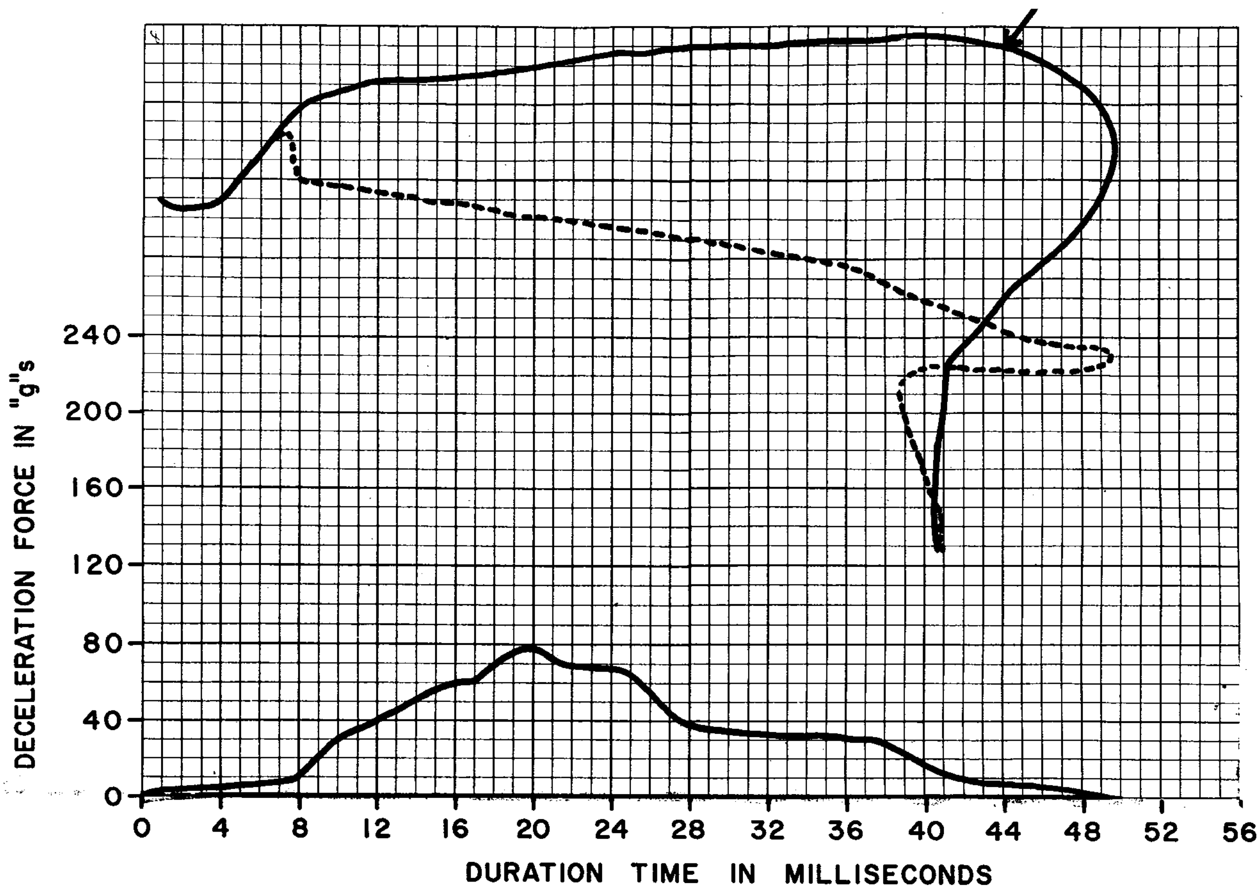


FIGURE 7. Zygomatic arch and nose impact. Both fractured.



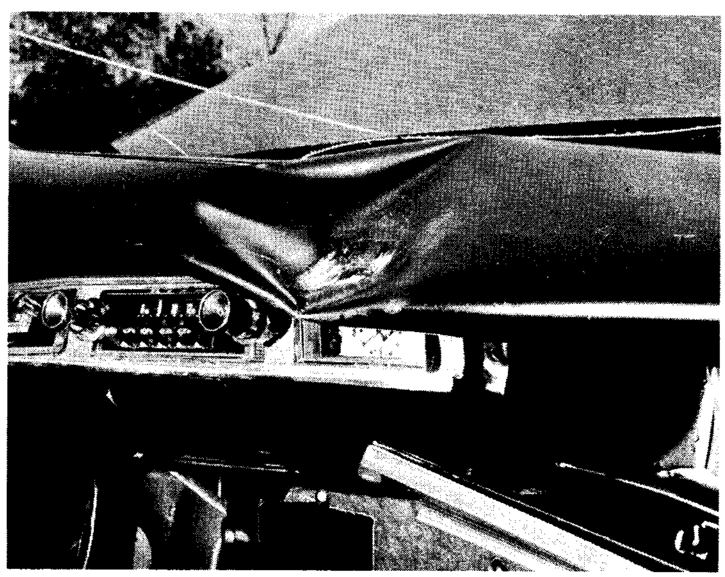
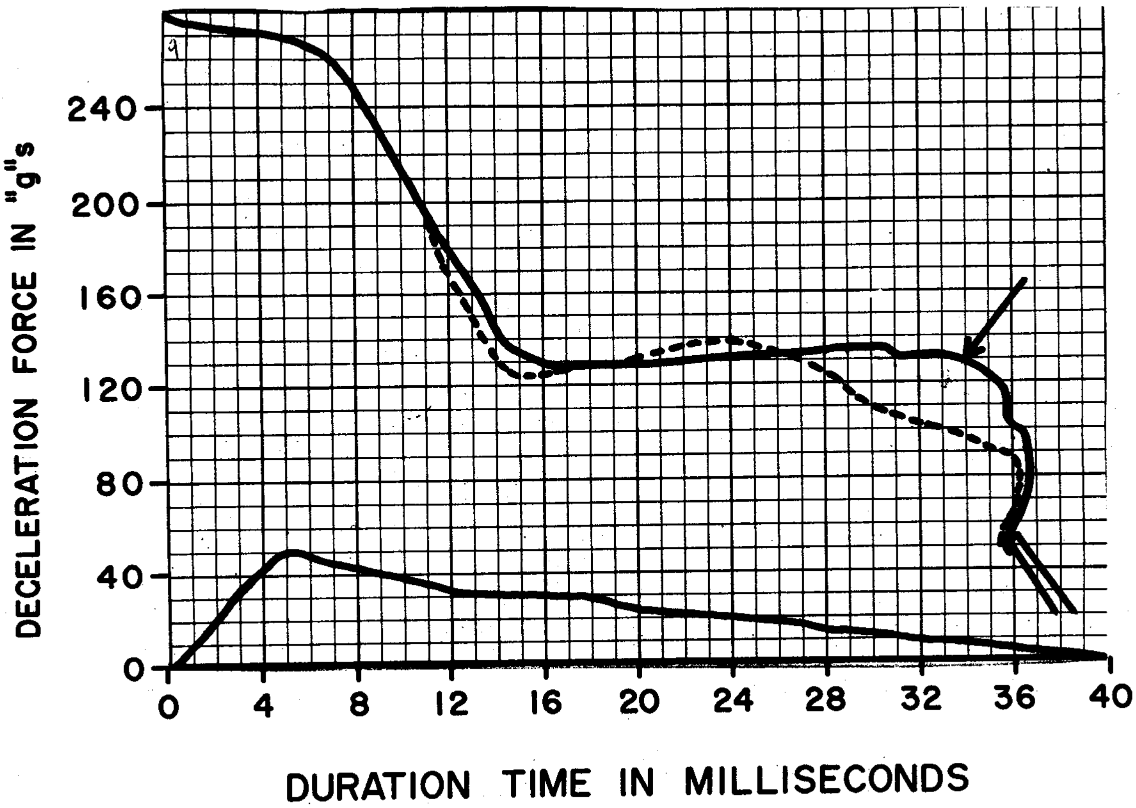


FIGURE 8. Impact of chin, teeth, and maxilla (note teeth marks). All received crushing fractures.



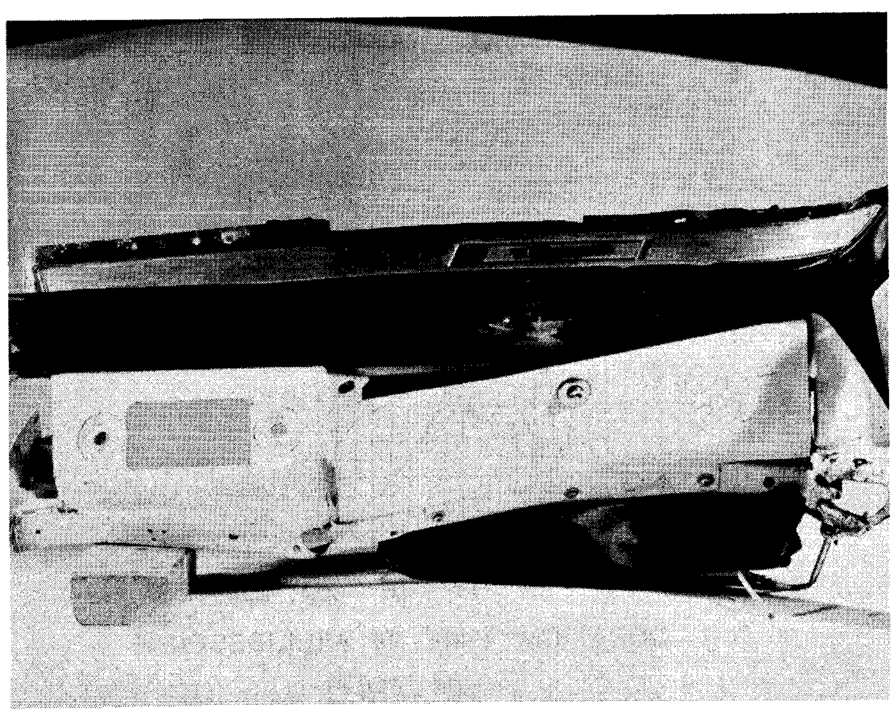
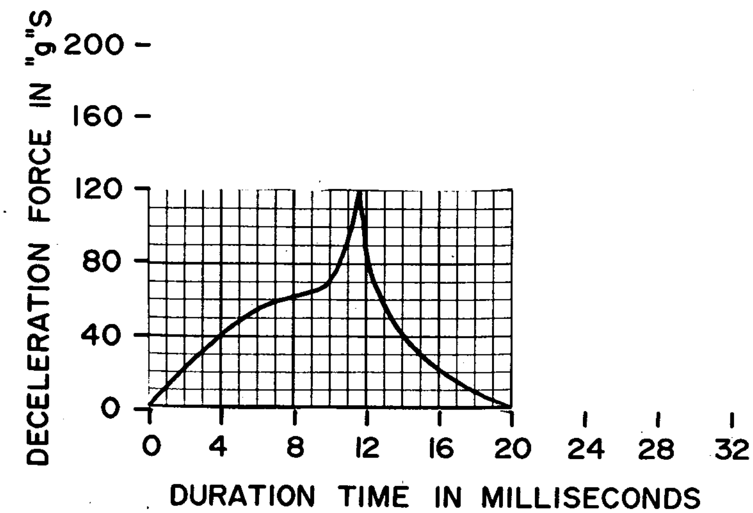
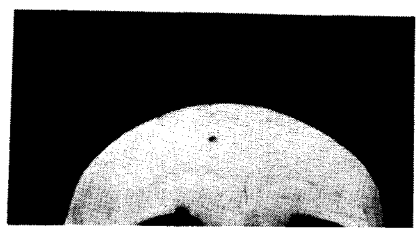
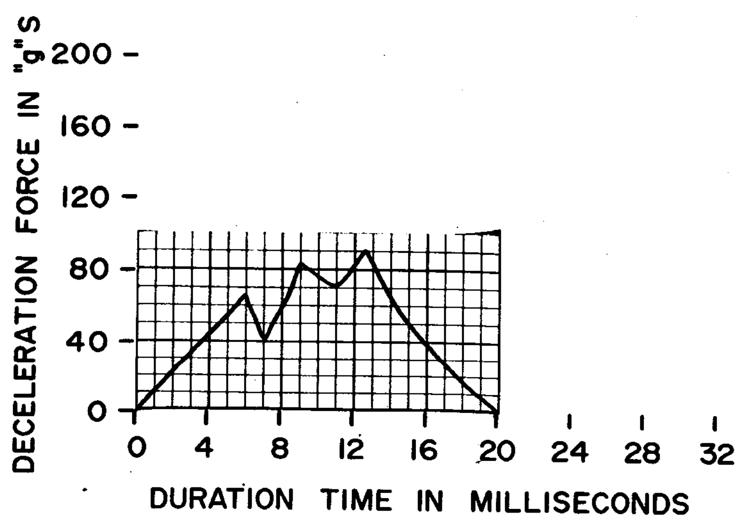


FIGURE 9. Impact of nose, both zygomatics, and upper teeth. No fractures, received bruises under eyes and concussion.



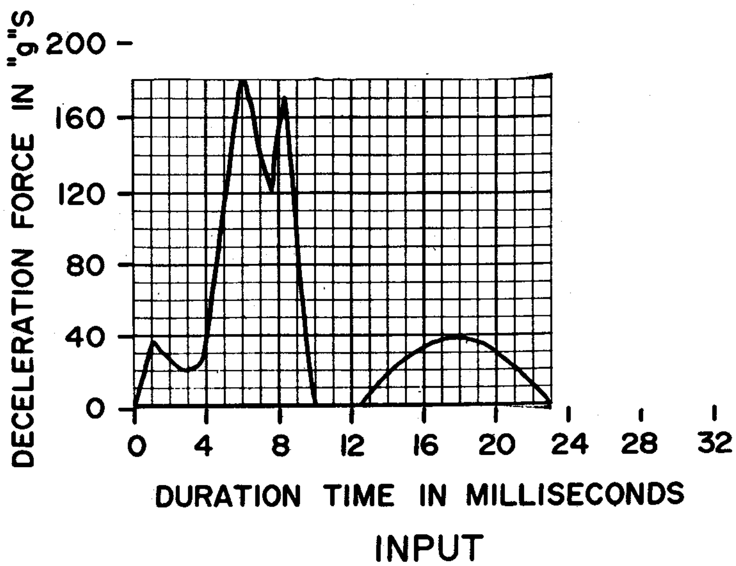
INPUT





OUTPUT

FIGURE 10. Cadaver-head impact against 1-inch-square block on forehead against deformable metal. Velocity of impact—20.88 ft/sec, stopping distance—2⅜ inches. Hairline fracture.



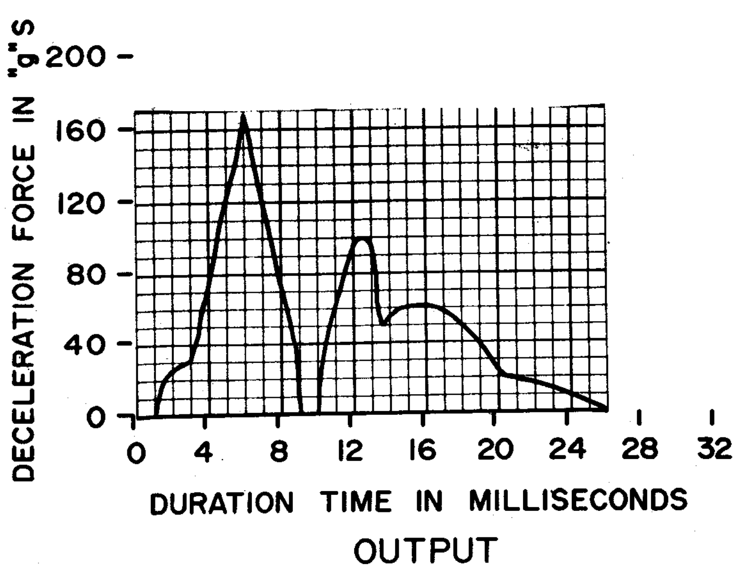
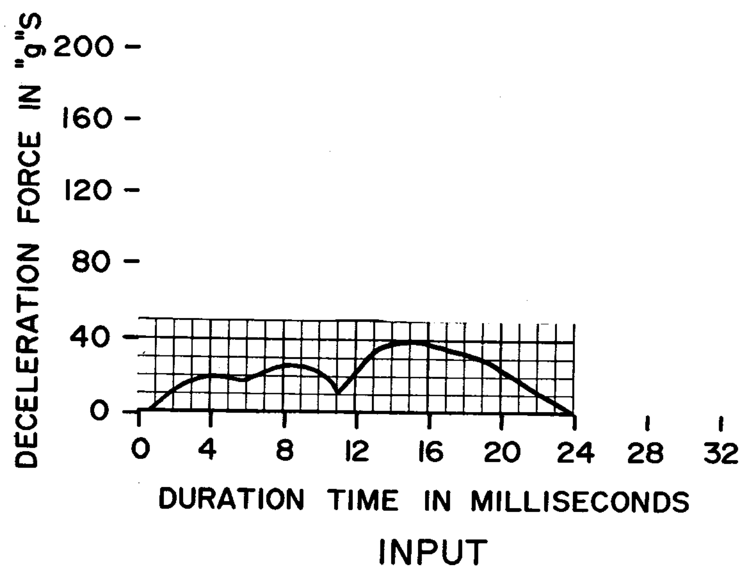


FIGURE 11. Crushing fracture with 1-inch-square block. Velocity—35.19 ft/sec, stopping distance—1⅞ inches.



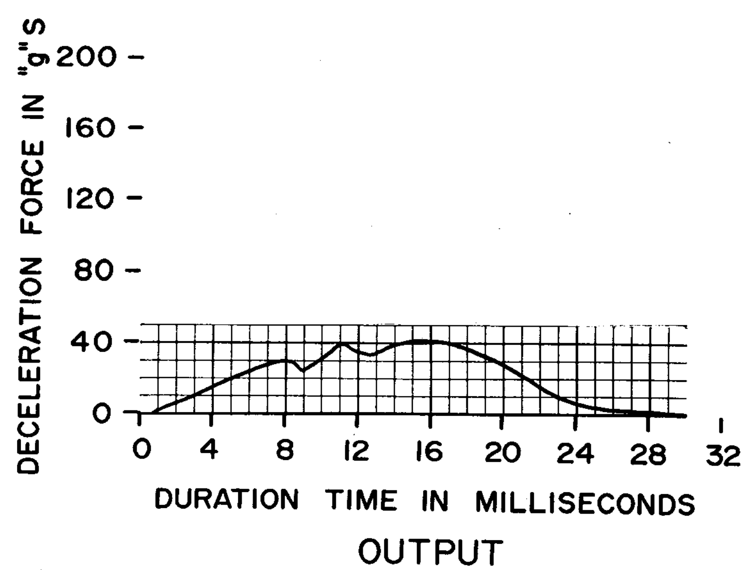
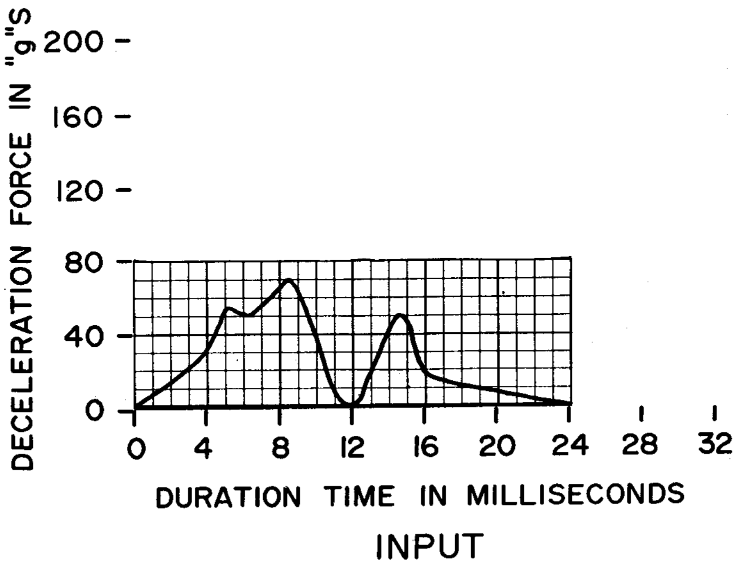
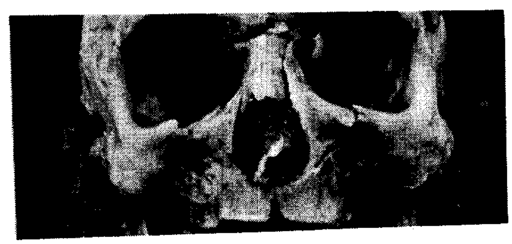


FIGURE 12. Hairline fractures of nose-zygomatic connections from a 12-ft/sec impact on molded block on the nose. Stopping distance—1½ inches.





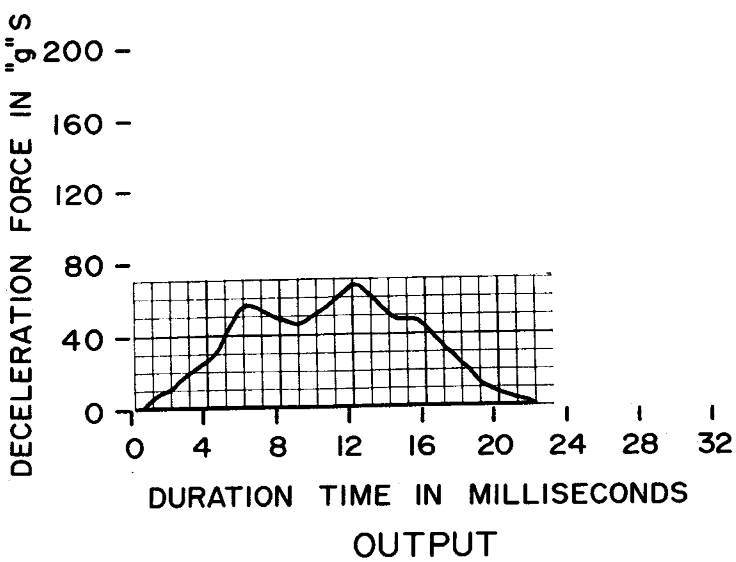


FIGURE 13. Crushing nose fracture. Velocity—14.81 ft/sec, stopping distance—⅝ inch.

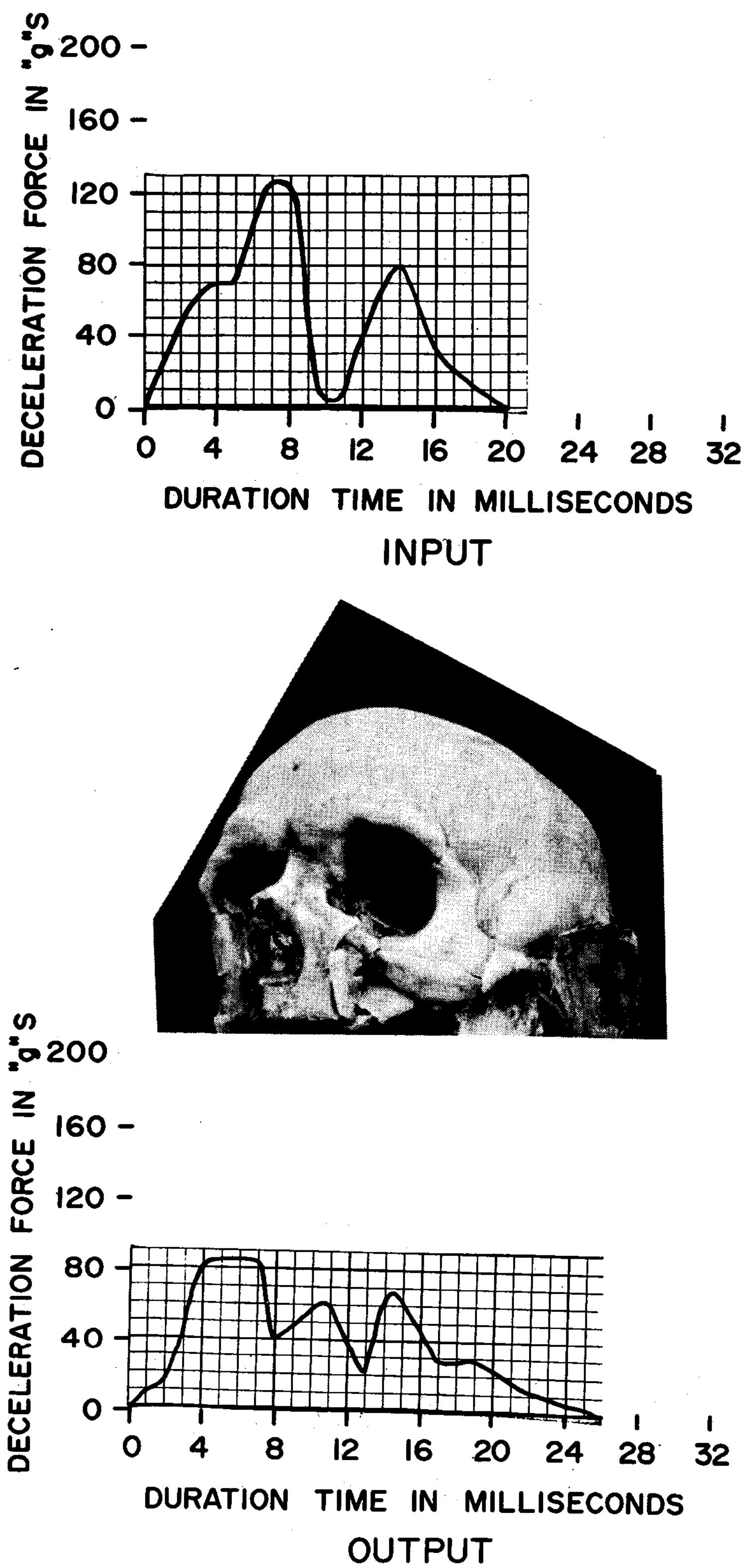


FIGURE 14. Both zygomatic arches crushed inward. Velocity—25.8 ft/sec, stopping distance—1 inch.

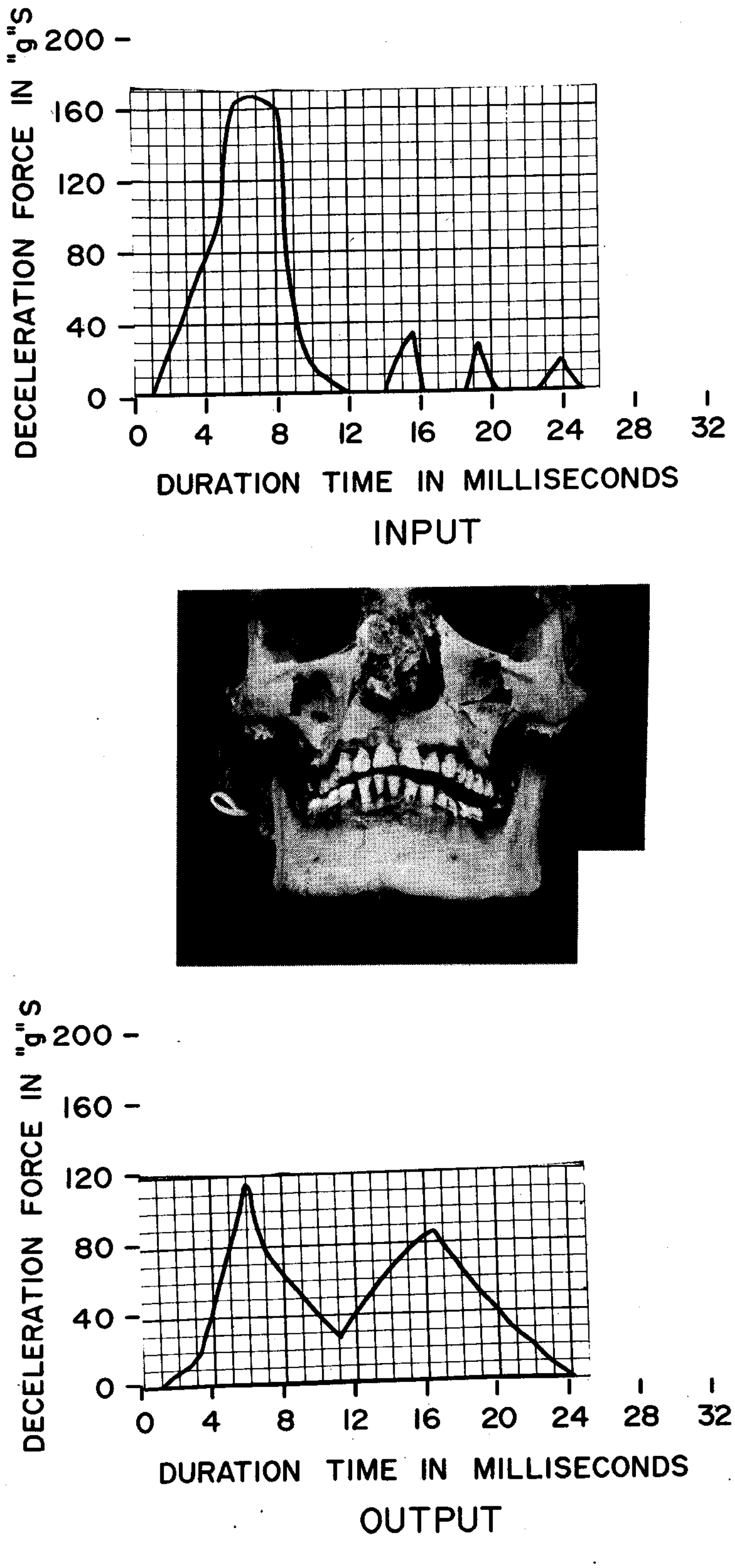


FIGURE 15. Fractures produced by 3.6-sq. in. block molded to fit teeth and maxilla. Fractures of maxilla and compound fracture of mandible. Velocity—35.97 ft/sec, stopping distance—1-3/16 in

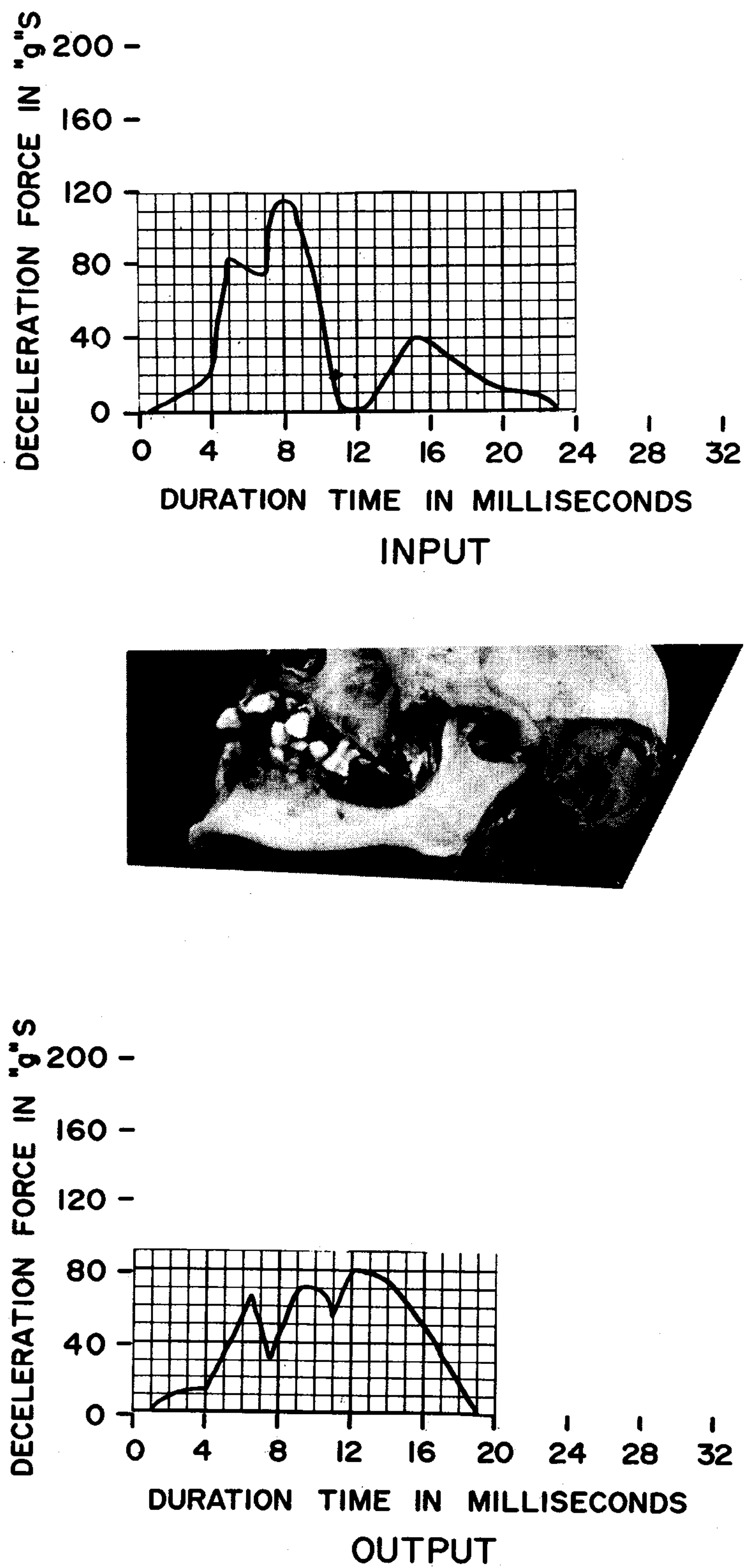


FIGURE 16. Typical mandibular condyle fracture from impact on molded block on tip of chin. Velocity—29.23 ft/sec, stopping distance—1 inch.

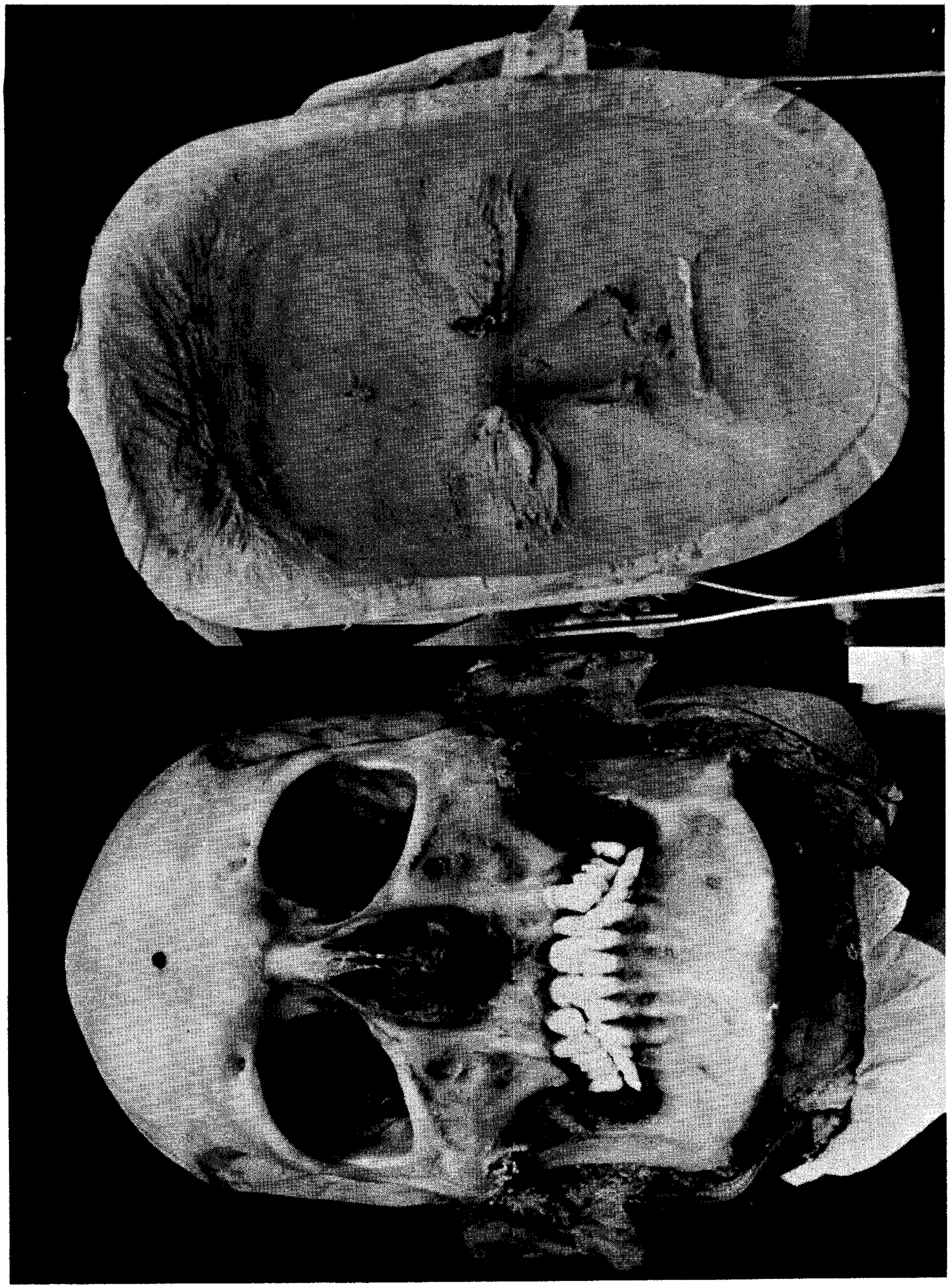


FIGURE 17. Cadaver head impacted at over 300 g with a full-face mask block. Complete lack of injuries.

FIGURE 18. Group of impact blocks used in this study.

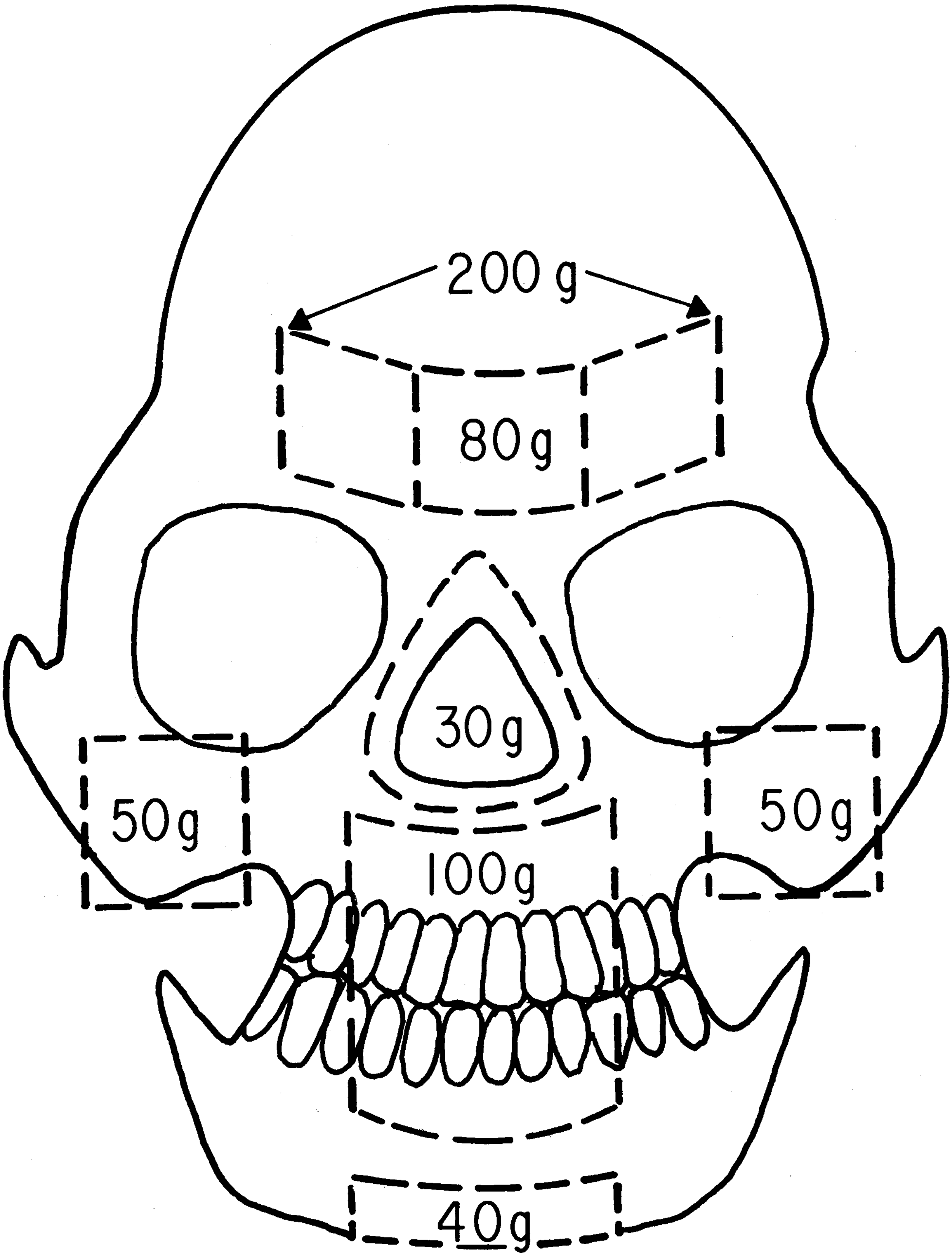


FIGURE 19. Summary of maximum tolerable impact forces on a padded deformable surface.

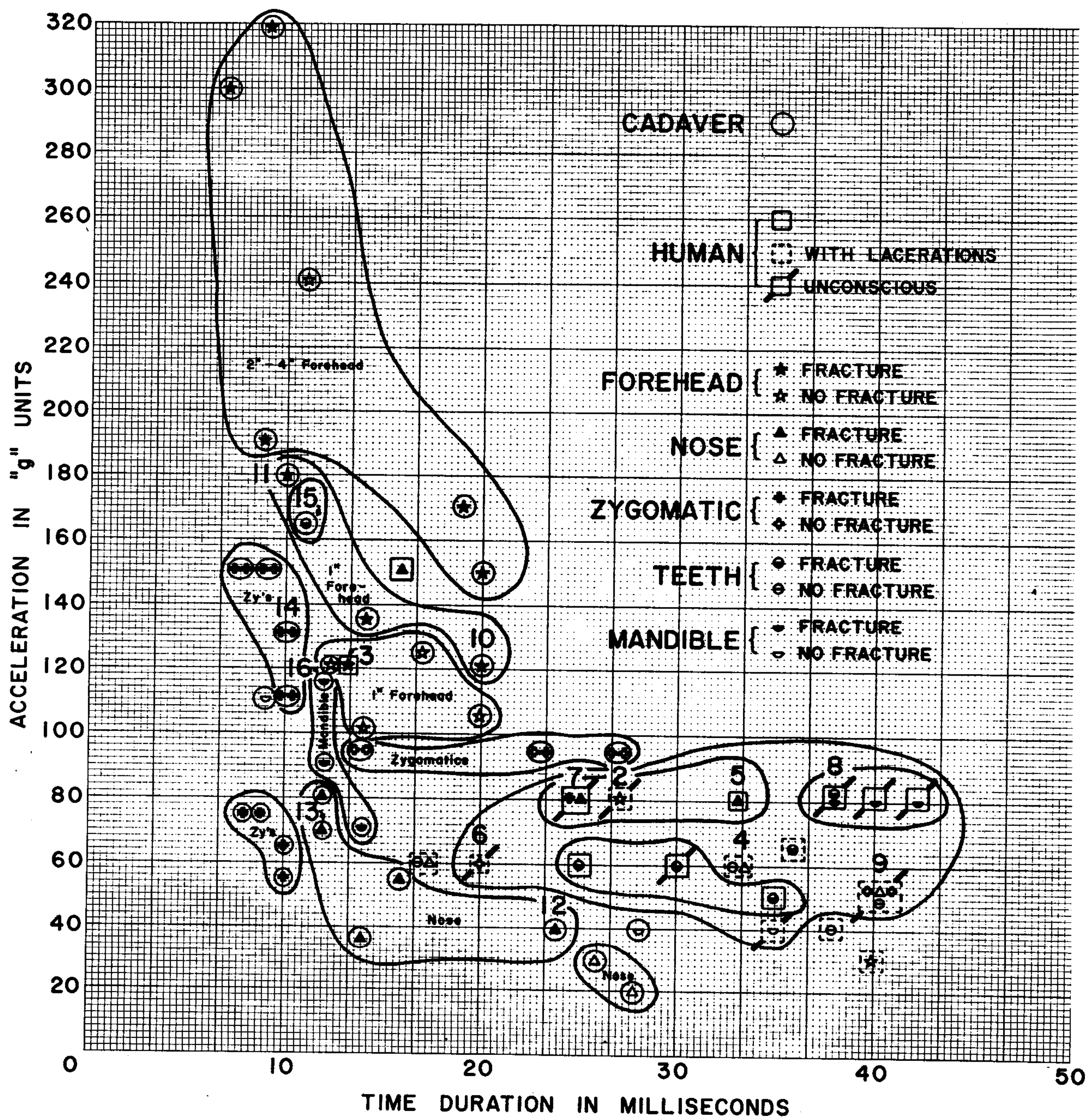


FIGURE 20. Graphic summary of human and cadaver facial tolerances to impact. Numbers correspond to figure numbers in this report.

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