EFFECTS OF HEAD RESTRAINT POSITION ON NECK INJURY IN REAR IMPACT

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Abstract

Although whiplash is typically classified as a minor injury, the economic cost of whiplash has been estimated at roughly $4.5 billion per year in the United States. International research efforts have included computational modeling, crash reconstructions, biomechanical testing of animals and human cadavers, and experimental rear impact tests using both anthropomorphic dummies and human subjects. Despite these efforts, the mechanisms and tolerances of whiplash injuries are still largely unknown.

This paper uses a computational modeling approach to better understand the effects of head restraint position on the risk of cervical injury under rear impact conditions. Both height and backset are varied over a wide range, while various engineering parameters believed to be related to cervical injury risk are examined. Simulation results for this model of a 50th percentile male indicate that for a head restraint height of 3 1.5 inches, the relative rotations, relative displacements, tensile loads, shear loads, bending moments, N_{TE} values and NIC values are relatively low for backsets of about 2 inches or less. Similarly, the results indicate that for a head restraint height of 29.5 inches, the value of these parameters are relatively low for backsets of l inch or less. These lower values suggest that a head restraint position that is higher and closer to the head may lead to a reduced risk of cervical spine injury.

Introduction

Whiplash injuries of the neck are one of the most common injuries reported from automotive rear impacts. Although these injuries are classified as minor (AIS I), their high incidence rate and often long-term consequences lead to significant societal costs. Based on data collected in the National Automotive Sampling System Crashworthiness Data System (NASS CDS), it has been estimated that more than 740,000 whiplash injuries occur annually[l]. Assuming an average cost of more than $6,000 per injury (including medical, legal, insurance, loss of productivity, and loss of work), the total annual monetary cost of whiplash in the United States is roughly $4.5 billion.

Since January 1, 1969, passenger cars have been required by Federal Motor Vehicle Safety Standard (FMVSS) No. 202 to provide head restraints that meet specified requirements for each designated

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front-outboard seating position. In 1991 this standard was extended to light trucks and vans, multipurpose passenger vehicles (MPVs), and buses with a gross vehicle weight rating (GVWR) of 10,000 pounds or less. The standard requires that either of two conditions be met:

1.) During a forward acceleration with a maximum value of 8.0 to 9.6 g over a duration of 80 to 96 msec on the seat supporting structure, the rearward angular displacement of the head reference line shall be limited to 45E from the torso reference line; or

2.) The head restraint must measure at least 27.5 inches above the seating reference point, with the head restraint in its fully extended position. The width of the head restraint, at a point 2.5 inches from the top of the head restraint or at 25 inches above the seating reference point, must not be less than 10 inches for use with bench seats and 6.75 inches for use with individual seats. The head restraint must withstand an increasing rearward load until there is a failure of the seat or seat back, or until a load of 200 pounds is applied. When the load is such that the applied moment is 3300 inch-pounds, the portion of the head form in contact with the restraint must not exceed a rearward displacement (perpendicular to the extended torso reference line) of 4 inches.

Condition 2 is almost universally used by the automotive industry, leading to a head restraint design based on geometric position rather than dynamic performance.

In 1982, NHTSA estimated the effectiveness of head restraints in reducing the overall risk of injury in rear impacts at 17% for integral head restraints and 10% for adjustable head restraints [2]. The effectiveness of adjustable head restraints may be lower most likely because they are frequently left in the down position.

Whiplash injuries were originally thought to be caused by hyperextension of the neck as the head rotated rearward over the seat back. However, recent studies by McConnell *et al.* [3] reported that some healthy middle-aged male subjects exposed to low speed rear impact of 4-8 kph experienced transient, mild cervical strain without exceeding the normal voluntary range of motion. Despite numerous studies being conducted on human volunteers, cadavers, and animals, no consensus has been reached on this difficult issue although several new theories have been proposed. Bogduk *et al.* [4] have isolated pain from whiplash to the facet capsules. Ono et *al.* [5] have observed that torso ramping causes compressive loading on the cervical spine, causing the lower vertebral segments to undergo motions beyond the normal physiological range. Svensson et *al.* [6] have investigated the effects of localized flexion and extension on the fluid pressure within the spinal canal. Common symptoms of whiplash injury include neck pain, headaches, blurred vision, tinnitus, dizziness, concussion and numbness[7]. Some of these symptoms are consistent with damage to the cervical muscles, ligaments and vertebrae while others are more difficult to explain since there are no lesions present on x-ray, CT scan or MRI.

In addition to head restraint height, several studies have investigated the effect of backset on neck injury during rear impact. Backset is defined as the distance in the horizontal plane between the occiput of the head and the head restraint. A study by Volvo showed that a significant increase in injury occurred when the occupant's head was more than 4 inches away from the head restraint[8]. Another study by Volvo reported that when vehicle occupants had their heads against the head restraint during impact, no injury occurred [9]. Thus, some manufacturers are pursuing automatic head restraint positioning systems which are capable of adjusting to the >optimum position by moving both vertically and horizontally. However, since there is currently no consensus on the mechanisms of whiplash injury, it is impossible to determine the maximum backset which still minimizes neck injury.

This paper summarizes the results of an analytical study using MADYMO [10]simulations to investigate the effects of head restraint position, including both height and backset, on the risk of cervical injury in rear impact conditions.

Methods

The MADYMO dynamic simulation program was used to study the effect of head restraint height and backset on several dependent variables associated with neck injury during rear impact. The baseline model of the standard Hybrid III dummy for the 50th percentile male was modified to include properties of the human neck, which will be referred to as the de Jager model. The de Jager neck model consists of rigid bodies for the head (CO), the cervical vertebrae (C1-C7) and the first thoracic vertebra (T1). The inertial characteristics of the head and neck are lumped into these nine rigid bodies. The rigid bodies are connected through three-dimensional nonlinear viscoelastic intervertebral joints which represent the lumped mechanical behavior of the intervertebral disc, ligaments, facet joints and muscles [11]. Thus, the neck joints are permitted six-degrees of freedom with joint characteristics that were derived from the experimental behavior of motion segments of the upper and lower cervical spine. The multibody neck-model was validated using the head-neck responses of human volunteers subjected to sled accelerations simulating frontal impact. Overall, the model satisfactorily matched the linear acceleration, trajectories of the occipital condyles and center of gravity of the head, angular acceleration, and head rotation of the human volunteers. However, the head rotations in forward flexion were greater than that for the human volunteers, especially at flexion angles greater than 50 degrees. Since the objective of this study is to determine the effects of head restraint position on cervical injuries in rear impact, these differences in flexion behavior between the human volunteers and the de Jager neck were judged not to be significant to these analyses.

The de Jager model was then positioned in a MAYDMO **multi-body** seat model previously developed by the University of Virginia[12]. The model was based upon a production seat and included a seat back and seat bottom which was made up of ten and thirteen segments, respectively. The seat characteristics modeled included the seat back hinge stiffness, seat back cushion energy return, head restraint cushion stiffness, head restraint cushion energy return, seat back hinge energy return, and friction coefficients. The seat back cushion and head restraint cushion energy return quantities, specified as 5%, were used as unloading functions for the MADYMO hysteresis model. Thus, the unloading force-penetration curve for the cushion is only 5% as stiff as the loading function. Similarly, the seat back hinge energy return was 67%. The friction coefficient between the Hybrid III dummy and the seat back was specified to be 0.7, while the friction coefficient between the head and the head restraint was specified to be 0.5. The seat back was modeled at a 20 degree recline from the vertical. The head restraint dimensions were about 6 inches in height, 12 inches in width and 5 inches in depth. In the original model developed by **WA,** the head restraint was parallel to the seat back plane. However, to better match the design of most head restraints, the head restraint was repositioned parallel to the vertical plane. Due to restrictions in MADYMO, the seat back and seat bottom are not rigidly connected but instead are joined by a special system of point restraints which act like a spring to resist rotation of the seat back. The load curves for the point restraints were defined to allow a maximum seat back rotation of about 5 degrees for the 18 kph pulse used in this study. This value is in agreement with various experimental studies on rear automotive impact. For instance, the maximum seat back rotation of a production seat for a 198 1 and 1982 Ford Escort with test subjects weighing as much as 93 kg during a 16 kph rear impact was less than 10 degrees[13]. A similar study using six production seats with a Hybrid III 50" percentile male dummy showed dynamic seat back rotations ranging from 3 to 16 degrees during a 16 kph rear impact[14].

Based on an intermediate value for the allowable maximum acceleration and pulse duration for the dynamic head restraint test specified in FMVSS 202, a 9 g sinusoidal acceleration pulse with a 90 millisecond duration was modeled (Figure 1). This acceleration pulse produced a maximum change in velocity of 18 kph which can be considered a low-speed collision.

A total of 21 MAYDMO simulations were performed to study the effect of head restraint height and backset on neck injury. Three head restraint heights were studied corresponding to the height specified by FMVSS 202 of 27.5 inches, a higher height of 3 1.5 inches which is specified in European regulation (ECE25), and a third height midway between the others at 29.5 inches. At each height, backset positions ranging from 0 to 6 inches in 1 inch increments were modeled. The forces, moments, and angular displacements **were** calculated at each of the eight vertebral joints. However for brevity of presentation, only the forces and moments at the atlanto-occipital joint (the joint between the head and the first cervical vertebra, C 1) are presented. By the sign convention used[15], positive shear forces are associated with translation of the head rearward with respect to the cervical spine, positive axial forces indicate tension, and positive moments indicate flexion of the cervical spine. The relative rotation and relative resultant displacement between the head and the first thoracic vertebrae, T1 , was also calculated by integrating the resultant relative accelerations. FMVSS No. 202 specifies for the dynamic test option that the maximum allowable rearward angular displacement of the head reference line with respect to the torso reference line is 45 degrees.

The newly proposed whiplash injury criterion, NIC, was also investigated. Based on the hypothesis by Aldman[16], Bostrom et *al.* proposed that the hydrodynamic pressure change in the spinal canal during maximal retraction was responsible for the soft tissue neck injuries [17]. The parameter NIC is given by the formula,

$$\mathrm{NIC} = 0.2\, a_{rel} + v_{rel}^{2} \qquad (1)$$

where a_{rel} is the resultant relative acceleration between first thoracic vertebra (Tl) and first cervical vertebra (Cl), v_{rel} is the resultant relative velocity between T1 and C 1, and the constant 0.2 represents the length of the cervical spine for human in meters. Corresponding to the point where the cervical spine reverses its natural curvature as the head moves rearward, the NIC_{50} value was proposed to be calculated at 50mm of relative T1 -Cl displacement with a proposed tolerance level of 15 m^2/s^2. The NIC_{50} and maximum NIC values were calculated in this paper to examine the trend of NIC values with respect to the changes of backset positions and head restraint heights.

Using the newly proposed Nij neck injury criteria by NHTSA[18] the probability of injury was compared for the 21 simulations. Nij combines the effects of forces and moments measured at the occipital condyles and is expected to provide a better predictor of craniocervical injuries than individual components of force and moment. The calculation of the Nij criteria yields four distinct injury risk values, namely tension-extension (N_{TE}), tension-flexion (N_{TF}), compression-extension (N_{CE}), and compression-flexion (N_{CF}). However, since head restraints are designed to limit the tension forces and extension moments on the neck during whiplash injuries, this report will focus on the values for N_{TE} in tension-extension. For the case of tension-extension injuries, the axial tension load (F_Z) is normalized with respect to a critical value for tension (F_{crit}). Extension moment (M_{ext}) is similarly normalized with respect to a critical value for extension (M_{crit}). Critical values for calculating the Nij are uniquely defined for each specific dummy size. The normalized neck tension-extension criteria can be written as the sum of these two normalized loads.

$$N_{TE} = (F_Z/F_{crit}) + (M_{ext}/M_{crit}) \qquad (2)$$

The normalizing critical values used for neck injury were 3600 N for tension and compression, 410 N-m for flexion, and 125 N-m for extension based on the 50th percentile adult male. These critical values were developed for the Hybrid III 50th-percentile dummy and are not directly applicable to a human neck. Because the Nij critical values were chosen to represent moderate to severe neck injuries at a value of Nij = 1, it is not clear what value of N_{TE} would best predict the probability of whiplash injuries. Thus, the results presented in this paper are viable only for looking at trends in the data, and not for quantitatively assessing the risk of injury.

Results

Tables 1 and 2 summarize the results for relative rotation and relative displacement of the head with respect to T1, tension, shear, and extension at the occipital condyles, NIC_{50}, maximum NIC and N_{TE}.

Kinematic Description

As expected, the kinematics of the head and cervical spine were very sensitive to the head restraint position, including both the height and backset. The head restraint height was important in determining both the location of the contact of the head restraint with the head and the direction of the force applied to the head. For instance, at the lowest head restraint height of 27.5 inches, the kinematics can be generally described in the following steps: (1) the head and torso translated backward together; (2) the top of head restraint contacted the head at the base of the skull, pushing the head upwards; and (3) the head rotated over the top of head restraint while the torso rebounded from the seat back cushion (Figure 2a). It should be noted that in the graphical representations, the kinematics are described from a reference frame which is attached to the vehicle seat bottom. At the intermediate head restraint height of 29.5 inches, the kinematics can be similarly described in the following steps: (1) the head and torso translated backward together; (2) the top of the head restraint contacted the occiput of the head, thereby limiting backward rotation of head; and (3) the head was pushed forward and upward while the torso rebounded forward (Figure 2b). At the highest height of 3 1.5 inches, the kinematics can be generally described in the following manner: (1) the head and torso translate backward together; (2) the center of head restraint contacted the occiput of the head, thereby limiting backward rotation of the head; and (3) the head was pushed forward while the torso rebounded (Figure 2c). Because the head restraint height determined the location of head contact with the head restraint and the direction of the force applied to the head, the head restraint height was of primary importance in determining the kinematics of the head and cervical spine during automotive rear impact.

The kinematics of the head and cervical spine were also sensitive to the head restraint backset. At a given head restraint height, the backset governs two main parameters: (1) the timing of contact between the head and head restraint relative to the contact between the torso and the seat back; and (2) the amount Of relative translation and rotation between the head and the upper torso. The backset positions analyzed in these simulations, which ranged from 0 to 6 inches in 1 inch increments, can be grouped into three categories that behave similarly - small, intermediate, and large backsets. At smaller backsets of about 0 to 2 inches, the head contacted the head restraint while the upper torso continued to move backward into the seat cushion. Consequently, the head restraint pushed the head forward with respect to the upper torso and there is minimal rearward rotation of the head with respect to the upper torso (Figure 3a). At an intermediate backset of about 2 to 4 inches, the head and torso rebounded from the head restraint and seat back at about the same time, which minimized the relative translation and rotation between the head and upper torso (Figure 3b). For larger backsets of about 4 to 6 inches, the upper torso began to rebound before the head contacted the head restraint. Consequently at the larger backset positions, the head continued to translate and rotate backwards relative to the upper torso until contact with the head restraint occurred (Figure 3c). Thus based on kinematics alone, it would appear that small to intermediate backset positions which minimize relative translation and rotation of the head with respect to the upper torso may lead to lower incidences of cervical spine injuries.

Focusing specifically on the cervical spine extension injuries during rear automotive impact, the relative rotation and relative resultant displacement of the head with respect to the first thoracic vertebra, T1, was calculated (Figures 4,5). The MADYMO simulations demonstrated that within the first 100 msec the head rotated forward with respect to T1 for all simulations. This can be explained by the backward rotation of the upper torso and seat back during the initial moments of impact while the head remained in its initial position. For the low head restraint height, the relative angular displacement followed a trend of flexion for the first 100 msec and then extension for the next 50 msec. Overall, the maximum relative rotation for the lowest head restraint height was very high, ranging from 30 to 66 degrees. Furthermore, there were large peak relative displacements at all backset positions for the lowest head restraint position, ranging from 70 to 170 mm (Figure 6a). For the intermediate and high head restraint heights at smaller backsets, the cervical spine remained in flexion throughout the simulation. However at larger backsets, the cervical spine followed a trend similar to the low head restraint height, i.e., initial flexion followed by extension. The relative rotation was relatively low (<5 degrees) for the intermediate height at backsets of I inch or less and for the high head restraint heights at backsets of 3 inches or less. The maximum relative displacements were also comparable for the intermediate (0 to 150 mm) and high head restraint positions (0 to 120 mm) (Figures 6b and 6c).

Forces and Moments at the Atlanto-Occipital Joint

In general, the most prominent effect of an increase in the head restraint height was a substantial decrease in the axial tension force at the atlanto-occipital joint (Figure 7). As described previously, at the lowest head restraint height, the head restraint contacted the head at the base of the skull and pushed the head upward. This caused high axial tension forces which exceeded 1000 N for most backset positions. However, for the cases of large backsets of 5 and 6 inches, the head rotated backwards and contacted the head restraint above the occiput, thus producing a compressive force which resulted in a decrease in the maximum tension. At the intermediate head restraint height, the top of the head restraint contacted the occiput of the head and pushed the head forward and slightly upward, resulting in lower maximum axial tension forces ranging from about 400 to 1700 N. At the high head restraint height, the center head restraint contacted the occiput of the head and pushed the head forward rather than upward. This produced the lowest maximum axial tension forces, ranging from about 100 to 1000 N. Thus, the overall trend seen for the three head restraint heights studied was decreasing axial forces with increasing head restraint height.

Although changes in the backset position affected the axial forces at the atlanto-occipital joint, the shear forces showed a stronger dependence on the backset position (Figure 8). Due to the rotation of the head over the top of the head restraint for the low head restraint heights, there is a general trend of increasing positive shear with increasing backset which is dependent on the specifics of the contact between the head and restraint at each backset position. However, the effect of backset on the shear forces can be more easily related to the kinematics for the intermediate and high head restraint heights. For instance, at a backset of 3 inches for the intermediate head restraint height, the head and upper torso translated backward in unison between 0 and 50 msec (Figure 9). Between 50 and 80 msec, the head moved backward relative to the upper torso, generating positive shear forces at the atlanto-occipital joint. At about 80 msec, the head contacted the head restraint, pushed the head forward and caused a decrease in the shear force. Thus, an increase in backset caused delayed contact of the head with the head restraint and an increase in the peak positive shear force.

The maximum extension torque at the atlanto-occipital joint was sensitive to both changes in backset position and head restraint height (Figure 10). Because positive shear forces caused extension moments, the trends for the torque at the atlanto-occipital joint are similar to that for the shear forces, *i.e.,* an increase in extension moment with increasing backset position.

NIC Injury Criterion

The NIC_{50} and maximum NIC values were sensitive to both head restraint height and backset position (Figure 11). In general, the NIC_{50} and maximum NIC were comparable when evaluated during the primary event of the first 100 ms. At the lowest head restraint height, a relative displacement of 50 mm was reached at all backset positions. However, the relative displacement was less than 50 mm for backset positions of 2 inches or less for the intermediate height and 3 inches or less for the highest height. Thus, the NIC_{50} values for those cases are 0. At the low head restraint height, there were large relative accelerations and velocities and consequently large values of NIC_{50} due to the rotation of the head over the head restraint while the torso was stopped by the seat back. At the larger backset positions of 5 to 6 inches, the NIC_{50} values did not show significant variation for the three head restraint heights because the 50 mm relative displacement was reached (-85 ms) before the head contacted the head restraint (-100 ms). Overall, the NIC_{50} value was less than the proposed limit of 15 m^2/s^2 for only the intermediate and highest head restraint positions at a backset of 1 inch or less.

Neck Injury Criteria, Nij

The newly proposed neck injury criteria, Nij, was used to synthesize the shear forces, axial forces, and flexion/extension moments into four injury risk values. Since head restraints are designed to limit the tension forces and extension moments on the neck during whiplash injuries, this report will focus on the value for neck injury in tension-extension, N_{TE} (Figure 12). As predicted from the high axial forces associated with contact of the head restraint with the base of the skull, N_{TE} was high, greater than 0.48, for the lowest head restraint height of 27.5 inches at all backset positions. By contrast, N_{TE} was two to three times lower at both the intermediate and high head restraint heights. In general, N_{TE} increased with increasing backset and decreasing head restraint height (Figure 12), corresponding to an increase in the maximum tension, positive shear, and extension moments. One exception to this trend is that N_{TE} decreased for the cases of 6 inches of backset primarily due to a decrease in the axial tension component. However, recall that the kinematics for the 6 inch backset case were not reasonable and showed large relative displacements and relative rotations. The intermediate head restraint height had relatively low (<0.2) values of N_{TE} when the backset was about 0 inches. Similarly, the highest head restraint height had relatively low values of N_{TE} at backset positions of 2 inches or less.

Discussion

Based on analyses using a model of a 50th percentile male Hybrid III anthropomorphic surrogate with human neck properties, this study demonstrates that an increase in head restraint height from 27.5 to either 29.5 or 31.5 inches above the seating reference point substantially reduced many of the engineering parameters believed to be related to the risk of cervical spine injury during low speed automotive rear impact. Furthermore these analyses demonstrated that decreasing the backset position also reduced the value of the relative displacements and loads on the cervical spine. Thus, assuming no other changes in seat design, the combination of a higher head restraint height and smaller backset position may provide the best protection from whiplash injuries during automotive rear impact. Simulation results for this model of a 50" percentile male indicate that for a head restraint height of 31.5 inches, the relative rotations, relative displacements, tensile loads, shear loads, bending moments, and N_{TE} values are relatively low for backsets of about 2 inches or less. Similarly, the results indicate that for a head restraint height of 29.5 inches, the value of these parameters are relatively low for backsets of I inch or less. The NIC_{50} criterion was somewhat more stringent, suggesting for this model that the backset should be 1 inch or less for both the intermediate and high restraint heights. However, this proposed value of 15 m^2/s^2 and the proposed constant of 0.2 m used in the formulation are subject to modification as further evaluation of the NIC_{50} criterion is completed.

These results agree well with other studies which demonstrate that increasing the head restraint height and reducing the backset position would reduce neck injury in rear automotive impact. First, a MADYMO simulation performed by the University of Virginia using a model of a Hybrid III with one neck segment also found that increasing the location of the center of the head restraint from 10 cm to 22 cm above the seat back resulted in a decrease in forces and moments at the head-neck joint[12]. In addition, a report by the Institute for Highway Safety rated a head restraint as "good" if both the distance from the top of the head down to the top of the head restraint was less than 6 cm (2.4 inches) and the backset distance was less than 7 cm (2.8 inches)[19]. By the year 2000, the European regulation will require that the front outboard seating positions to have a head restraint that can achieve a height of 31.5 inches above the seating reference point and have a minimum height at all outboard seating positions of 29.5 inches above the seating reference point. However, the European standard does not specify a backset requirement.

Although the de Jager model used in the current study was not validated for rear impact, our results show reasonable agreement with experimental results for the relative rearward rotation of the head with respect to the upper torso during low speed rear impact. For instance, Szabo *et al.* [13]demonstrated relative rotation from an initial position of about 25 degrees of flexion to a maximum of 10 degrees of extension for 6 test subjects who experienced a 16 kph (10 mph) rear impact. These subjects, who had seated heights ranging from 0.84 to 0.94 m (33 to 37 inches), were seated in a production seat with a 0.72 m (28.3 inch) seat back height. Based on these dimensions, the kinematics for the test subjects in the study by Szabo would probably resemble the kinematics described for the low or intermediate head restraint height in the current study. Thus, the maximum extension of 35 degrees from the initial position in the Szabo study is comparable to the 26 to 52 degrees relative rotation seen in the current study for the low head restraint height.

Our analyses demonstrated that the relative rearward rotation of the head with respect to the fit thoracic vertebrae was decreased by increasing the head restraint height and decreasing the backset position. In general, proper head restraint height and backset positioning which limits this relative rearward rotation should decrease the incidence and/or severity of cervical spine extension injuries. However, there is no consensus as to human tolerance to extension injuries. Originally, it was believed that hyperextension was needed to produce the classical whiplash symptoms. Previous studies on human volunteers have reported average normal range of motion of the head to be about 65 degrees in flexion and 75 degrees in extension[20]. The amount of relative rotation in the current study is well below this range for the high and intermediate head restraint heights at all backset positions. However, more recent studies by McConnell *et al.* [3] have suggested that hyperextension is not a prerequisite for whiplash injuries. In their study, four healthy male volunteers who ranged in age from 45 to 56 were subjected to a series of tests of rear impact at 12.9 kph (8 mph) or less. Despite the fact that none of these subjects exceeded their voluntary range of motion, three out of four volunteers experienced mild, transient cervical strain.

Although our MADYMO simulations attempted to carefully model the behavior of the neck through eight cervical spine segments with joint stiffness properties in three translational and three rotational directions, it should be cautioned that the results are meant to provide information about the general trends rather than exact values for the forces, moments, optimal positions, etc. The numerical results are sensitive to a number of parameters related to both the seat and the occupant. Since only one seat design was used for this study, factors such as the seat contour, head restraint geometry, cushion stiffness, head restraint stiffness, damping, and friction may influence the results. For instance, the combination of seat cushion and seat back which is “soft” dynamically and a head restraint that is positioned at a larger backset of 4 inches may perform just as well as a more rigid system with the head restraint positioned at 1 inch of backset. The study by University of Virginia addressed the effect of many seat design parameters and occupant size on the kinematics and mechanics of the neck using a model of a Hybrid III with a one-segment neck [12]. In another computer modeling study, seat design features such as plastic deformation in the seatback were shown to affect the likelihood of neck injury[21]. Further studies are needed to assess the influence of factors such as seat design parameters and occupant size on these results.

Since the current regulation for seat performance is a geometric and static strength test, the dynamic performance of the various seat designs in rear impact may result in varying amounts of seatback rotation. Typically, a bucket design front seat would experience more seat back rotation than the back seat of a sedan which is rigidly coupled to the car frame. An additionai parameter study to investigate the effects of an almost rigid seat back on whiplash injuries was performed. In the model, the seat rotational hinge stiffness was increased to limit the amount of seat back rotation to about 0.5 degrees. A total of 2 1 cases were analyzed with the same three head restraint heights (27.5", 29.5" and 3 1.5") and range of backsets (0"- 6" at 1" increments). For these cases of 0.5 degrees of seat back rotation, the kinematics were similar to those described for the baseline case of 5 degrees of seat back rotation. This led to similar trends of a decrease in the various neck injury parameters investigated with increasing head restraint height and decreasing backset.

Although only the 50th percentile male Hybrid III was modeled in this study, the results can be extended to adults of various sizes by analyzing sitting height relative to the head restraint height. Anthropometric data gathered for the development of the family of human surrogates determined that the average erect sitting heights for large males, mid-sized males and small females are 37.8, 36.4, and 32.0 inches, respectively [22]. The erect sitting heights were measured from the surface of a rigid seat to the top of the head, whereas the head restraint heights presented in this study are measured from the seating reference point which is located approximately at the hip joint. The difference in the sitting height of a little more than 1 inch suggests that a large male using the high head restraint height of 3 1.5 inches should have similar kinematics to a mid-sized male using the intermediate head restraint height of 29.5 inches. Further investigation is necessary to determine if a head restraint that is positioned too high for a small female may also lead to cervical spine injuries.

In summary, MADYMO dynamic simulations using a human surrogate the approximate size of a 50" percentile male with sophisticated neck properties demonstrated that an increase in head restraint height from 27.5 to either 29.5 or 3 1.5 inches above the seating reference point substantially reduced loads on the cervical spine and thus may reduce the risk of neck injury during a low speed rear impact. For a head restraint height of 3 1.5 inches, the relative rotations, relative displacements, tensile loads, shear loads, bending moments, N_{TE} values and NIC values are relatively low for backsets of about 2 inches or less. Similarly, the results indicate that for a head restraint height of 29.5 inches, the value of these parameters are relatively low for backsets of 1 inch or less. These lower values suggest that a head restraint position that is higher and closer to the head may lead to a reduced risk of cervical spine injury.

References

1. NHTSA. *Head Restraints - Identification of Issues Relevant to Regulation, Design and Effectiveness.* 1996. NHTSA Technical Report, NHTSA Docket 96-22-l. US Dept of Transportation. Washington, DC 20590.
2. *Kahane,* D.J. *An Evaluation of Head Restraint - Federal Motor Vehicle Standard 202. 1982.* NHTSA Technical Report DOT HS 806 108. National Technical Information Service. Springfield, VA 22161.
3. McConnell, W.E., R.P. Howard, H.M. Guzman, J.B. Bomar, J.H. Raddin, J.V. Benedict, H.L. Smith, and C.P. Hatsell. *Analysis of human test subject kinematic responses to low velocity rear end impacts.* SAE International Congress and Exposition. 1993. Detroit, Michigan, USA: Society of Automotive Engineers Inc. Warrendale Pennsylvania USA.
4. Bogduk, N. and A. Marsland, *The Cervical Zygapophyseal Joints as a Source of Neck Pain,* Spine, 1988. 13(6): p. 610-617.
5. Ono, K.. K. Kaneoka, A. Wittek, and J. Kajzer. *Cervical Injury Mechanism Based on the Analysis of Human Cervical Vertebral Motion and Head-Neck-Torso Kinematics During Low Speed Rear Impacts. 4* 1 st Stapp Car Crash Conference. 1997. Orlando, Florida, USA: Society of Automotive Engineers Inc. Warrendale Pennsylvania USA.
6. Svensson, M.Y ., B. Aldman, H.A. Hansson, P. Lovsund, T. Seeman, A. Suneson, and T. Ortengren. *Pressure Effects in the Spinal Canal during Whiplash Extension Motion: A Possible Cause of Injury to the Cervical Spinal Ganglia.* 1993 International IRCOBI Conference on the Biomechanics of Impact. 1993. Eindhoven, Netherlands.
7. Bogduk, N., *The Anatomy and Pathology of Whiplash. Clinical Biomechanics.* 1986. p. 92-l 0 1.
8. Olsson, I., 0. Bunketorp, G. Carlsson, C. Gustafsson, I. Planath, H. Not-in, and L. Ysander. *An In-depth Study of Neck Injuries in Rear-end Collisions.* International IRCOBI Conference on Biomechanical Impacts. 1990. Bron-Lyon, France.
9. Jakobsson, L., H. Norin, C. Jernstrom, S.E. Svensson, P. Johnsen, 1.1. Hellman, and M.Y. Svensson. *Analysis of Head and Neck Responses in Rear End impacts - A New Human-Like Model.* 1994. Volvo Car Corporation Safety Report.
10. MADYMO. TNO Road-Vehicles Research Institute. The Netherlands.
11. de Jager, M., A. Sauren, J. Thunnissen, and J. Wismans. *A global and a detailed mathematical model for head-neck dynamics.* 40th Stapp Car Crash Conference. 1996. Albuquerque, New Mexico, USA: Society of Automotive Engineers Inc. Warrendale Pennsylvania USA.

12. Sieveka, E.M.. K. Levent, and W.D. Pilkey. *Simulation of Occupant and Seat Responses in Rear Impacts.* 1995. Final Report, Contract NO. DTRS-57-93-C-00 105. US Dept of Transportation. Washington, D.C.

13. Szabo, T.J., J.B. Welcher, R.D. Anderson, M.M. Rice, J.A. Ward, L.R. Paulo, and N. J. Carpenter. *Human occupant kinematic response to low speed rear-end impacts.* SAE International Congress and Exposition. 1994. Detroit, Michigan, USA: Society of Automotive Engineers Inc. Warrendale Pennsylvania USA.

14. Prasad, P., A. Kim, D.P.V. Weerappuli, V. Roberts, and D. Schneider. *Relationships Between Passenger Car Seat Back Strength and Occupant Injury Severity in Rear End Collisions: Field and Laboratory Studies.* 4 1 th Stapp Car Crash Conference. 1997. Orlando, Florida USA: Society of Automotive Engineers Inc. Warrendale Pennsylvania USA.

15. SAEJ2 11. *Instrumentation for Impact Test.* 1995. Society of Automotive Engineers. Warrendale, Pennsylvania USA.

16. Aldman, B. *An Analytical Approach to the Impact Biomechanics of Head and Neck.* 30th Annual Proceedings of the Association for the Advancement of Automotive Medicine. 1986.

17. Bostrom, O., M.Y. Svensson, B. Aldman, H.A. Hansson, Y. Haland, P. Lovsund, T. Seeman, A. Suneson, A. Saljo, and T. Ortengren. *A new neck injury criterion candidate-based on injury findings in the cervical spinal ganglia after experimental neck extension trauma.* 1996 International IRCOBI Conference on the Biomechanics of Impact. 1996. Dublin, Ireland.

18. Kleinberger, M., E. Sun, R. Eppinger, S. Kuppa, and R. Saul. *Development of Improved Injury Criteria for the Assessment of Advanced Automotive Restraint Systems.* 1998. NHTSA Docket 98-4405-9. US Dept of Transportation. Washington, D.C.

19. IIHS. *Special Issue: Whiplash Injuries, Status Report.* 1995. Volume 30, No 8. Insurance Institute for Highway Safety. Arlington, VA 22201.

20. Buck, C.A., F.B. Dameron, M.J. Dow, and H.B. Skowlund, *Study of Normal Range of Motion in the Neck Utilizing a Bubble Goniometer.* Archives of Physical and Medicine Rehabilitation, 1959. 40: p. 390-392.

21. Nilson, G., M.Y. Svensson, P. Lovsund, and D.C. Viano. *Rear-End Collisions - The Effect of Recliner Stiffness and Energy Absorption on Occupant Motion.* 1994. ISBN 9 l-7 197-03 l-2. Dept of Injury Prevention, Chalmers University. Goteborg, Sweden.

22. Schneider, L.W., D.H. Robbins, M.A. Pflug, and R.G. Snyder. *Development of Anthropometrically Based Design Specifications for an Advanced Adult Anthropomorphic Dummy Family.* 1983. NHTSA Technical Report, National Technical Information Service. University of Michigan Transportation Research Institute. Springfield, VA 22 16 I.

Table 1: Summary of Results for de Jager Hybrid III 50th Percentile Male MADYMO Model

Height& Backset	Max Neg Relative Rotation (deg)	Time Neg Relative Rotation (ms)	Max Relative Disp, T1-C0 (mm)	Time Max Relative Disp (ms)	Max Tension (N)	Time Max Tension (ms)	Max Pos Shear (N)	Time Max Pos Shear (ms)	Max Ext Moment (N*m)	Time Max Ext Moment (ms)
Low 0	30.3	125	71.5	125	1284.5	120	111.2	80	12.8	95
Low 1	36.9	12s	103.6	130	1358.7	115	369.7	85	18.6	105
Low 2	43.9	125	116.3	130	1431.5	110	345.9	90	24.2	110
Low 3	46.8	130	128.3	130	1518.7	115	336.1	90	28.1	125
Low 4	47.4	130	138.7	115	1461.4	120	464.5	100	28.0	120
Low 5	54.2	130	153.6	120	1135.6	120	507.9	100	31.4	130
Low 6	66.2	135	168.3	125	872.8	115	509.4	100	36.3	135
Mid 0	0.0	0	1.4	50	416.9	80	3.6	35	5.7	90
Mid 1	0.0	0	16.1	80	889.7	120	3.6	35	9.8	96
Mid 2	9.3	115	41.8	90	1117.6	115	43.9	70	13.8	100
Mid 3	16.9	120	64.2	95	1444.9	115	154.3	60	16.2	105
Mid 4	19.9	120	88.9	100	1661.0	115	241.2	86	16.1	12s
Mid 5	24.7	120	111.1	105	1455.7	115	359.1	9s	20.4	120
Mid 6	45.0	130	146.0	115	928.6	146	506.9	100	24.5	125
High 0	0.0	0	1.6	50	127.6	90	3.7	35	1.3	70
High 1	0.0	0	14.0	80	163.9	155	3.7	35	3.1	95
High 2	0.0	0	36.1	90	108.6	150	40.6	70	6.0	100
High 3	1.7	120	55.8	95	510.1	110	129.3	80	9.3	105
High 4	9.7	115	76.3	100	838.2	115	212.3	85	12.8	120
High 5	15.3	115	99.2	105	976.2	120	329.6	90	16.8	120
High 6	28.5	125	116.7	110	500.5	115	419.8	95	18.2	120

Table 2: Summary of Results for de Jager Hybrid III 50th Percentile Male MADYMO Model

Height& Backset	NIC50 (m2/s2)	Time NIC50 (ms)	Max NIC (m2/s2)	Time Max NIC (ms)	Max Nte	Time Max Nte (ms)
Low 0	0.0	**	16.6	95	0.48	120
Low 1	0.0	**	21.6	70	0.58	115
Low 2	25.8	80	25.8	80	0.67	110
Low 3	35.5	85	35.5	85	0.73	120
Low 4	36.1	85	36.1	85	0.70	120
Low 5	36.1	85	36.1	85	0.61	125
Low 6	36.2	85	36.2	85	0.48	125
Mid 0	0.0	*	[illegible]	70	0.17	80
Mid 1	0.0	*	8.9	95	0.31	120
Mid 2	0.0	*	19.1	65	[illegible]	110
Mid 3	0.0		25.2	75	0.60	115
Mid 4	32.9	85	32.9	85	0.68	115
Mid 5	36.3	85	36.3	85	0.64	120
Mid 6	36.3	85	36.3	85	0.43	120
High 0	0.0	*	9.7	100	0.08	90
High 1	0.0	*	5.9	55	0.09	90
High 2	0.0	*	19.2	65	0.11	100
High 3	0.0	**	24.5	75	0.26	110
High 4	25.5	80	26.6	80	0.39	115
High 5	35.5	85	36.5	85	0.45	120
High 6	36.5	85	36.5	85	0.30	115

* Did not achieve 50 mm of relative displacement

** NIC50 values were negative at 50 mm of relative displacement

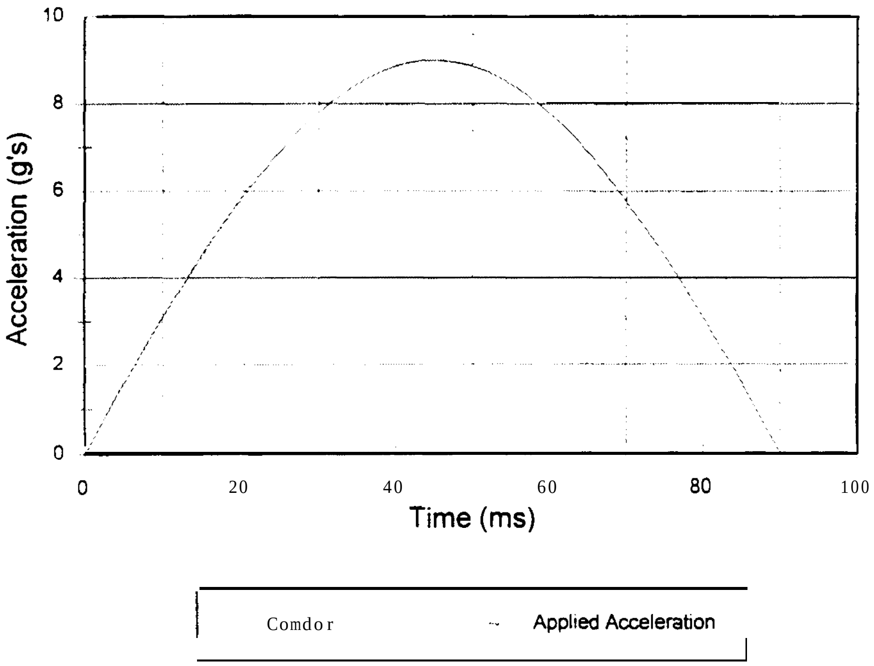
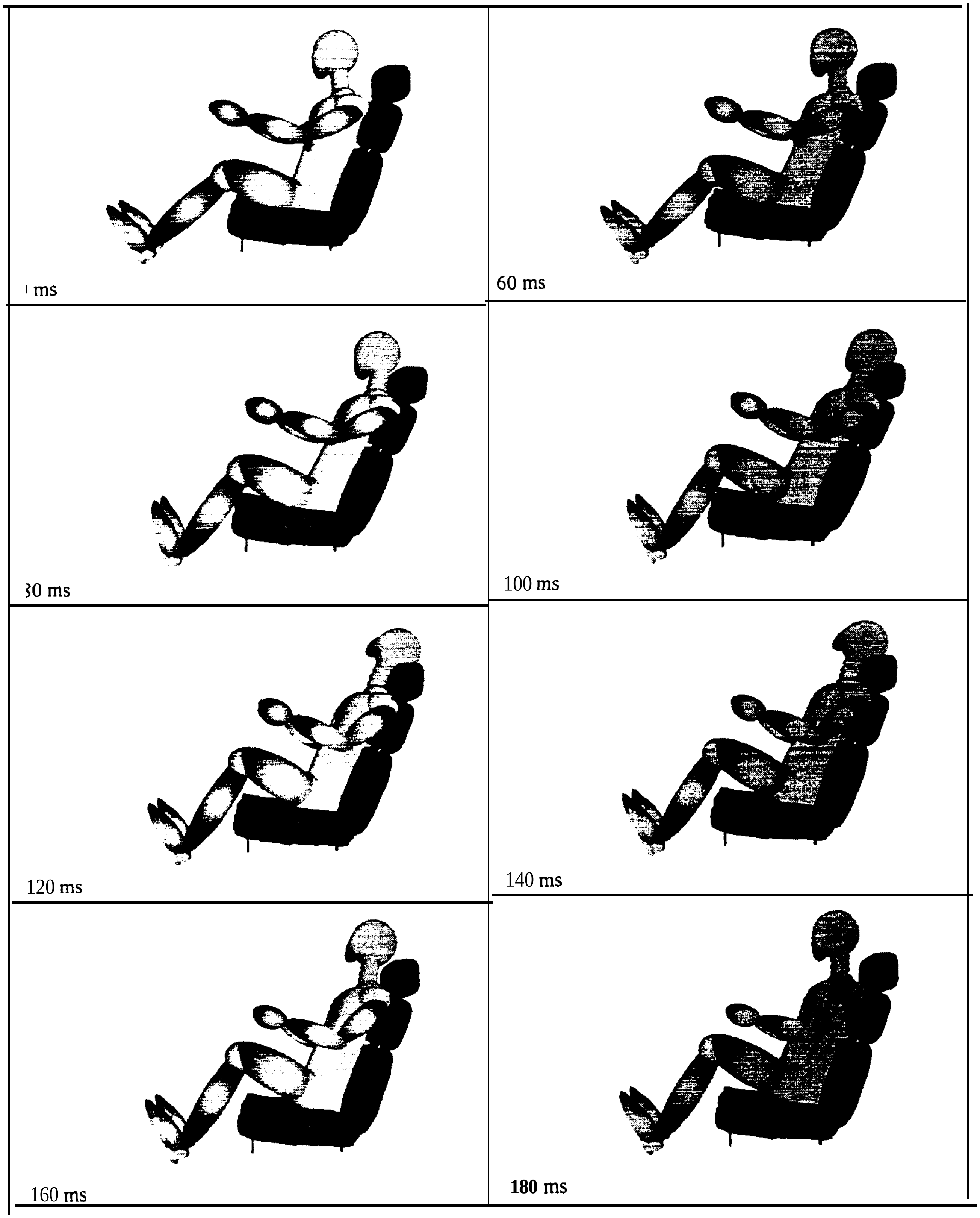


Figure 1: Applied acceleration pulse and FMVSS No. 202 specified corridors



ms

60 ms

0 ms

100 ms

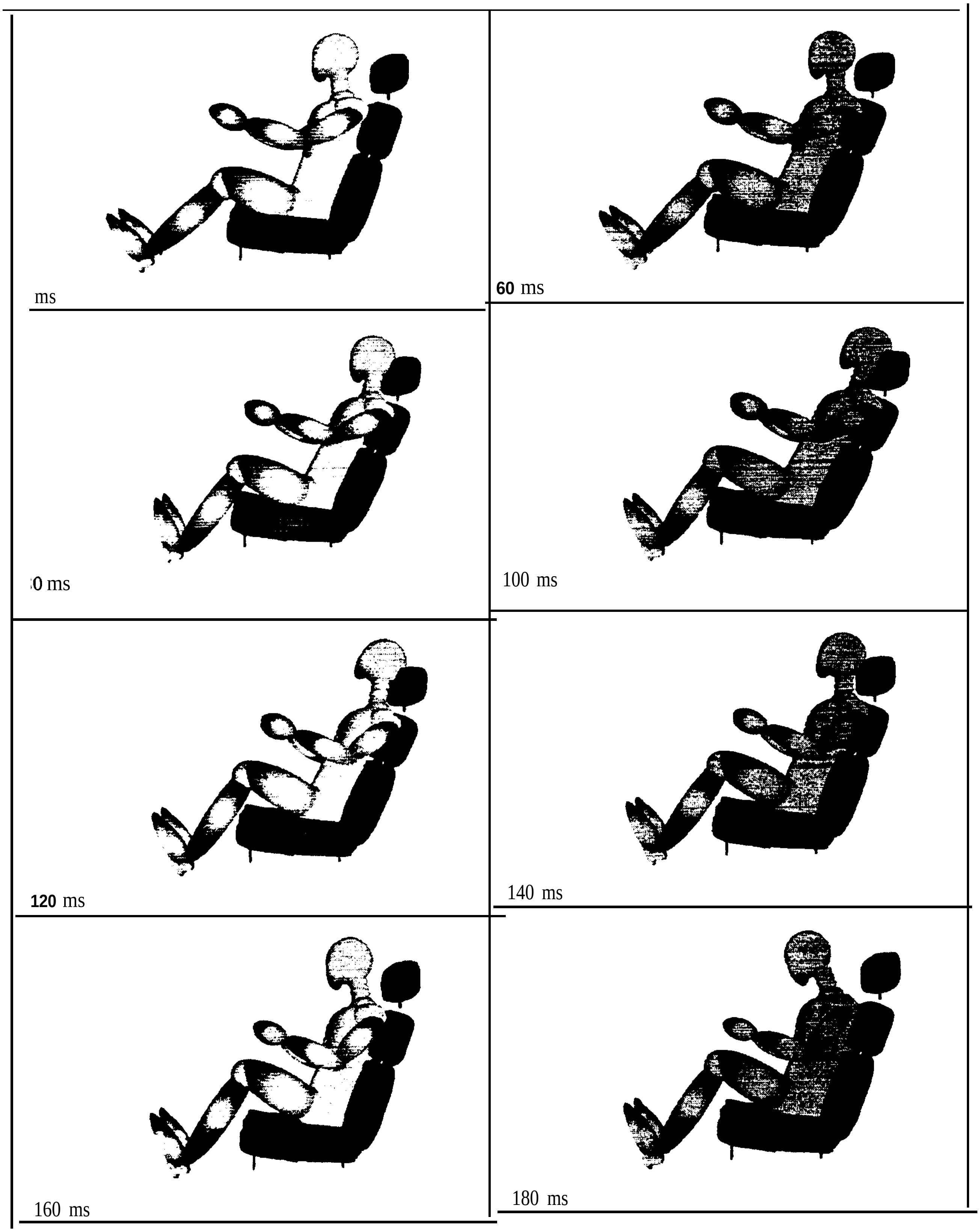
120 ms

140 ms

160 ms

180 ms

Figure 2a: Kinematics for Low Head Restraint Height = 27.5”, Backset = 3”



ms

60 ms

0 ms

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120 ms

140 ms

160 ms

180 ms

Figure 2b: Kinematics for Intermediate Head Restraint Height = 29.5", Backset = 3"

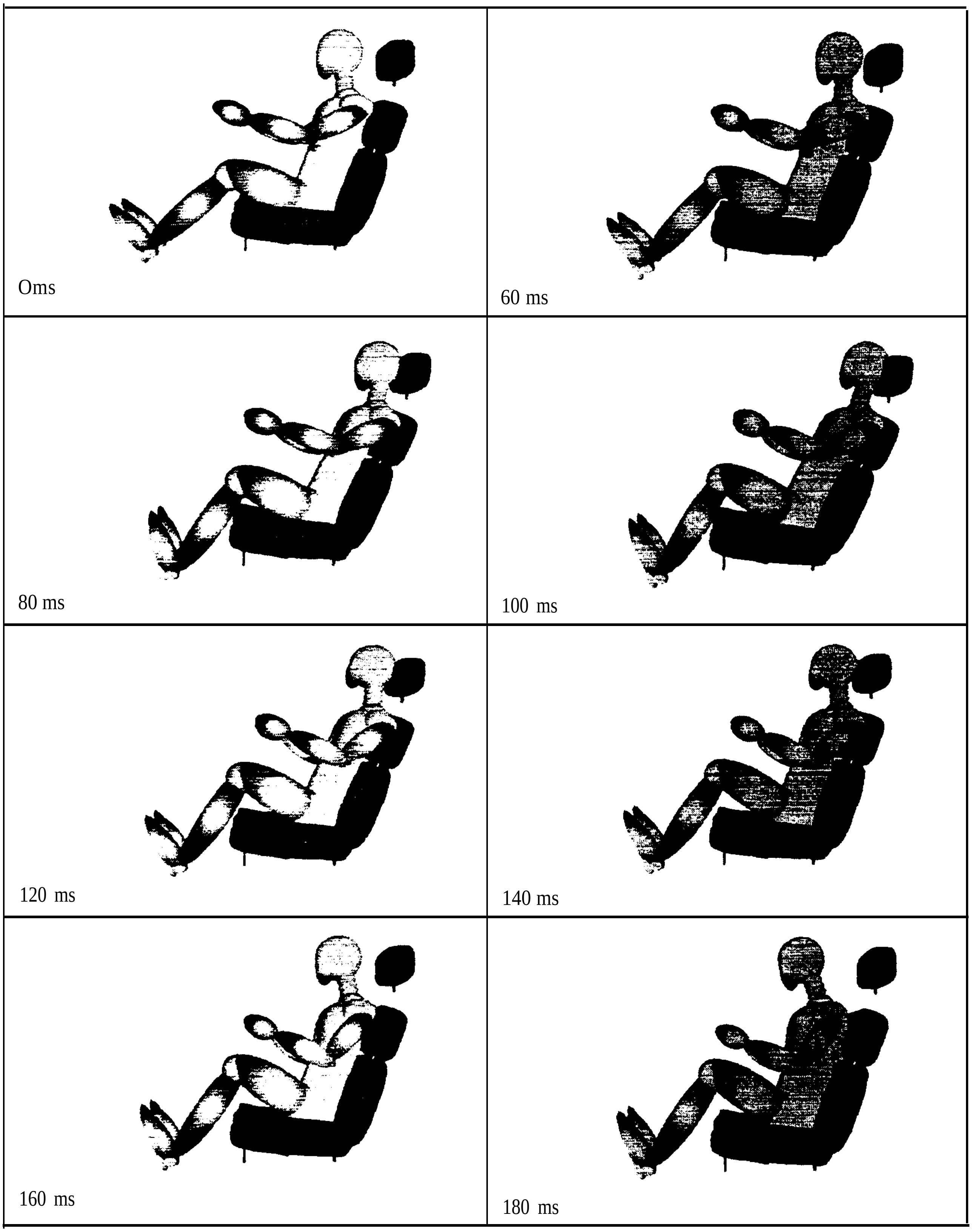
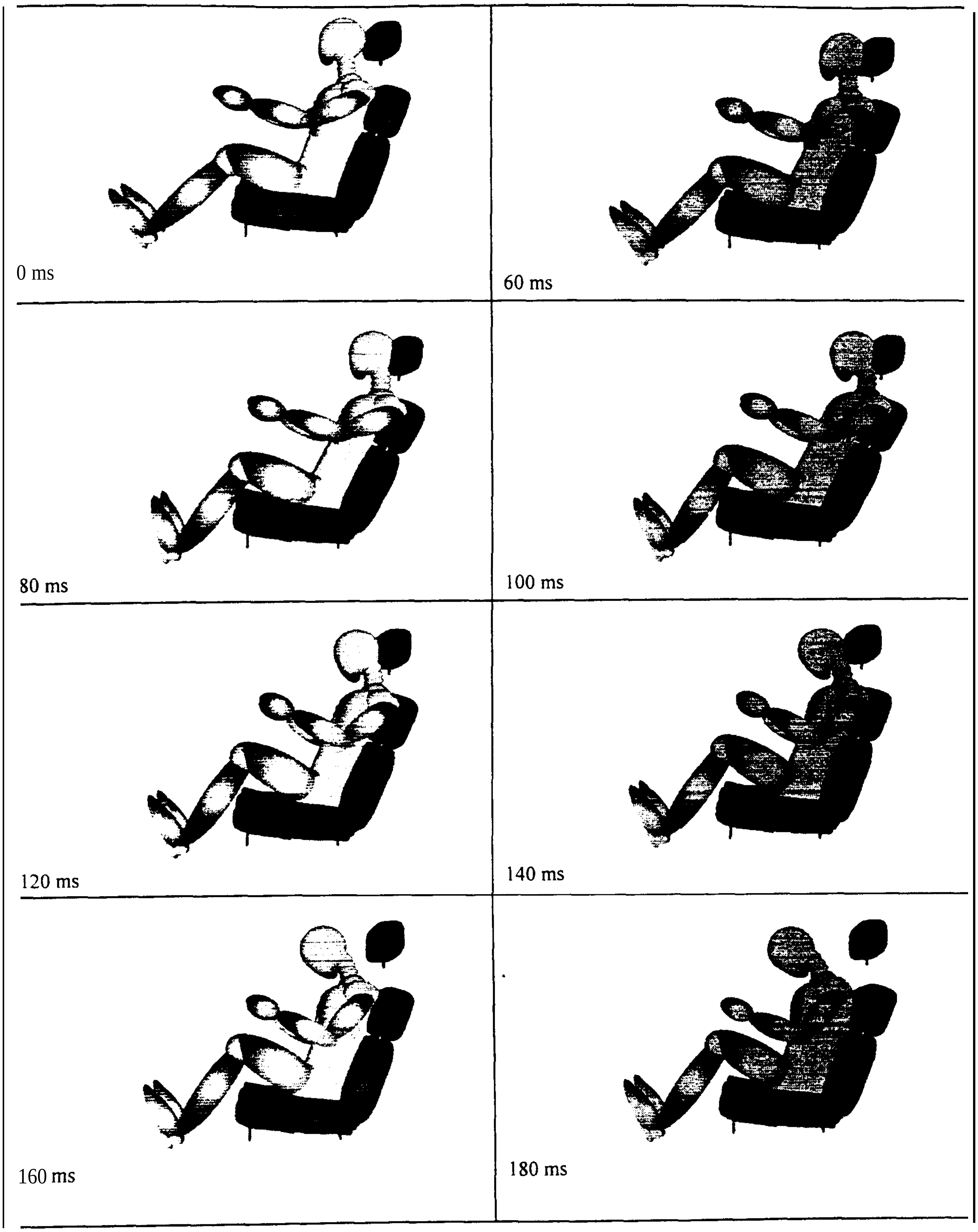


Figure 2c: Kinematics for High Head Restraint Height = 31.5", Backset = 3"


0 ms

60 ms

80 ms

100 ms

120 ms

140 ms

160 ms

180 ms

Figure 3a: Kinematics for High Head Restraint Height = 31.5", Backset = 0"

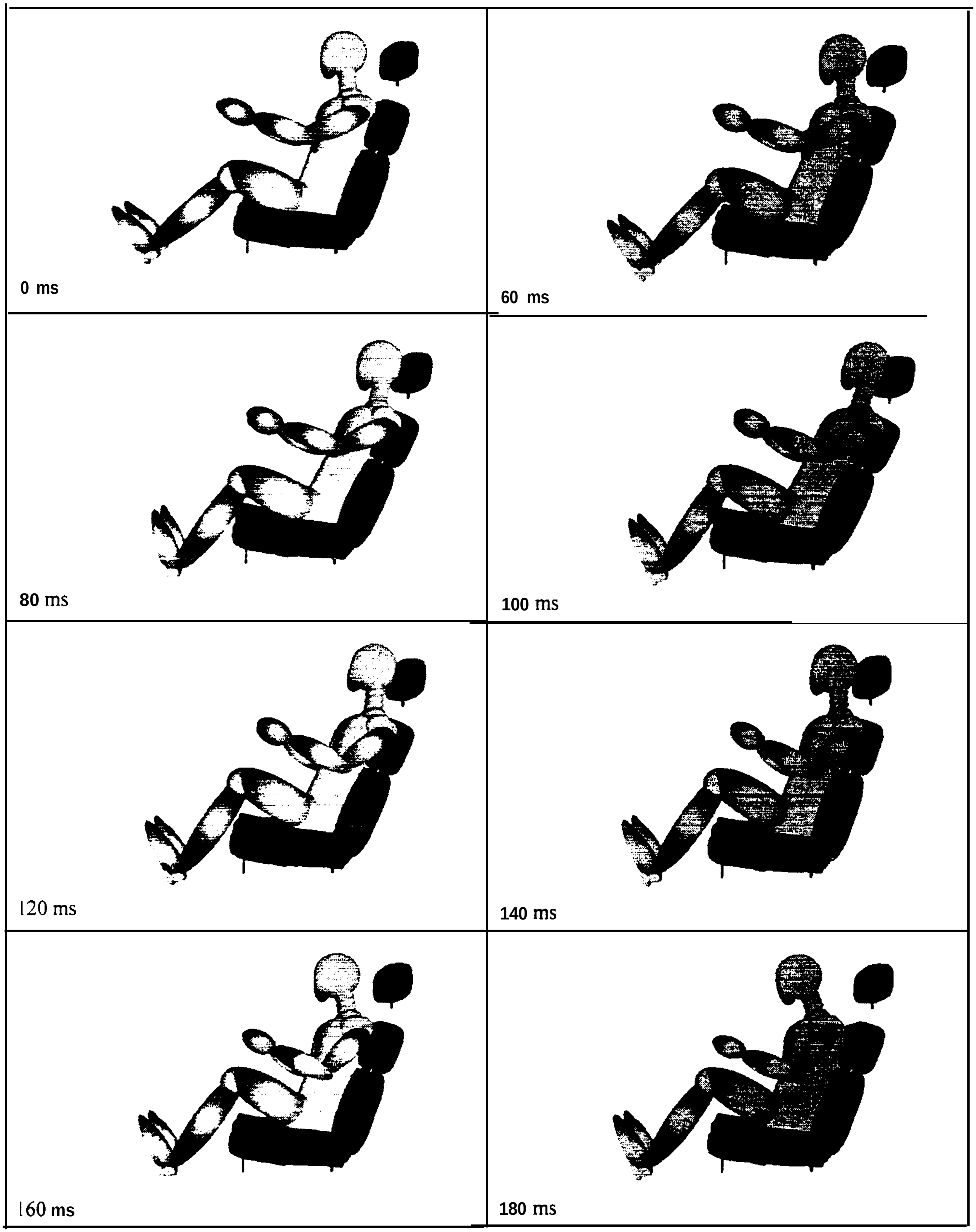


Figure 3 b: Kinematics for High Head Restraint Height = 31.5”, Backset = 3”

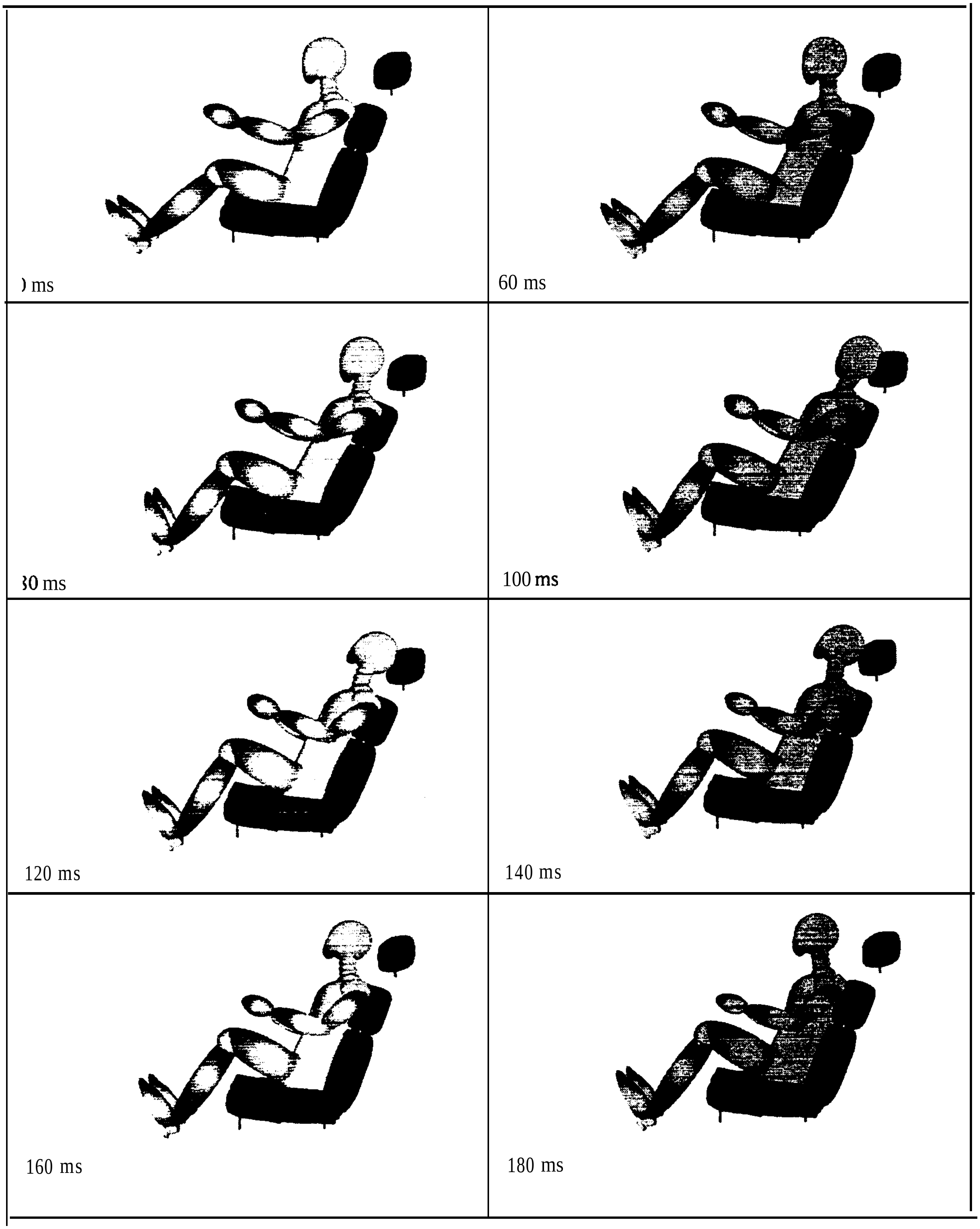


Figure 3c: Kinematics for High Head Restraint Height = 3 1.5”, Backset = 6”

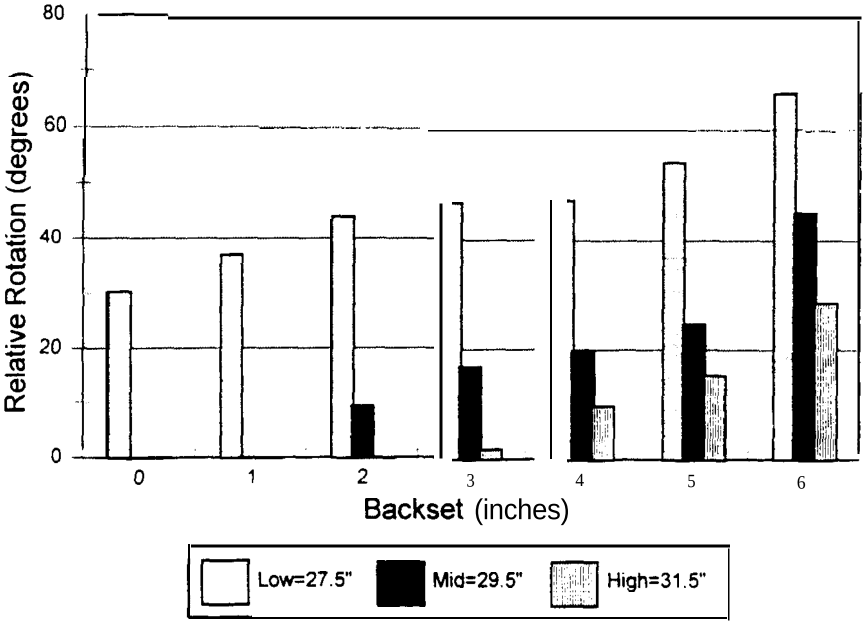


Figure 4: Maximum Relative Negative Head Rotation. The maximum rotations decreased with increasing head restraint height and decreasing backset.

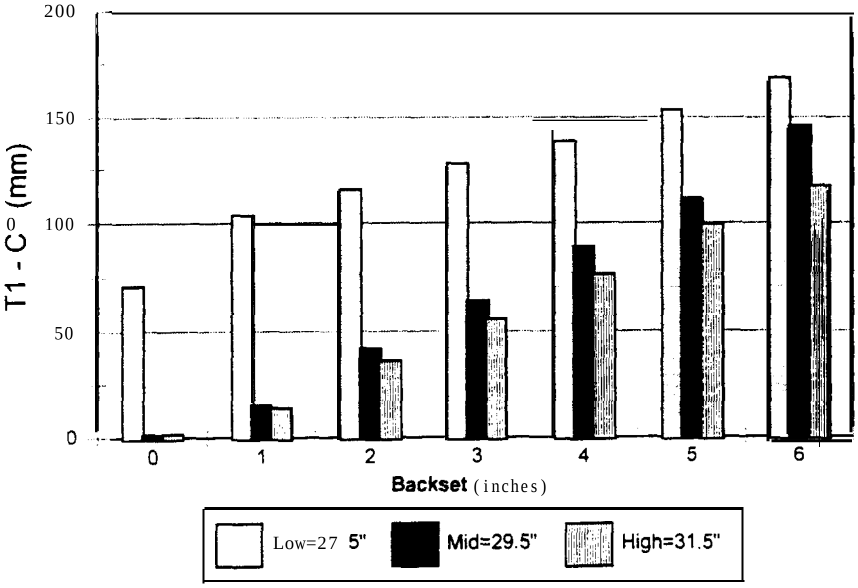


Figure 5: Maximum Relative Displacement. The maximum displacments also decreased with increasing head restraint height and decreasing backset.

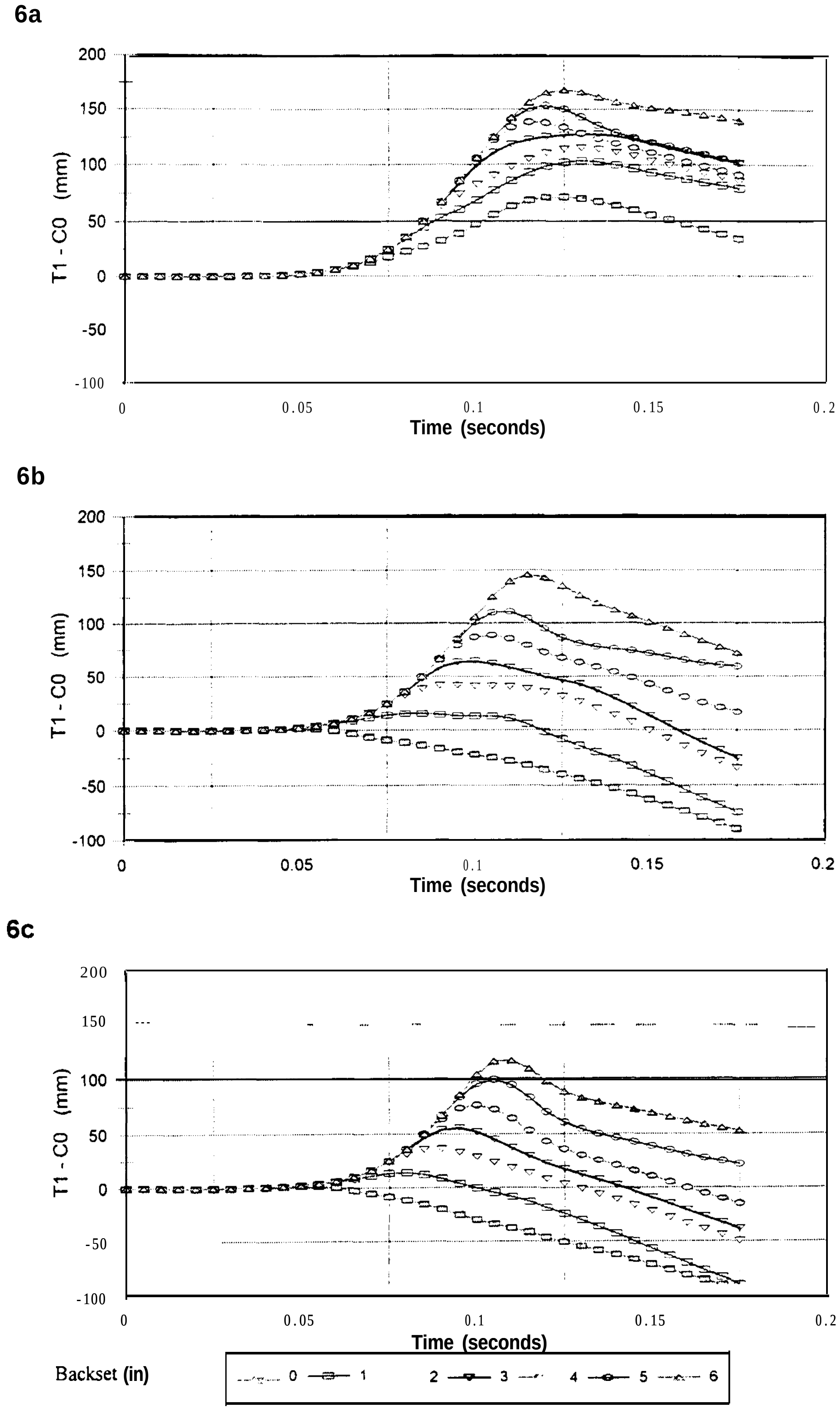


Figure 6: Relative Displacements of the Head for the Low(6a), Intermediate(6b) and High(6c) Head Restraint Positions. Positive values indicate rearward movement of the head relative to the thorax.

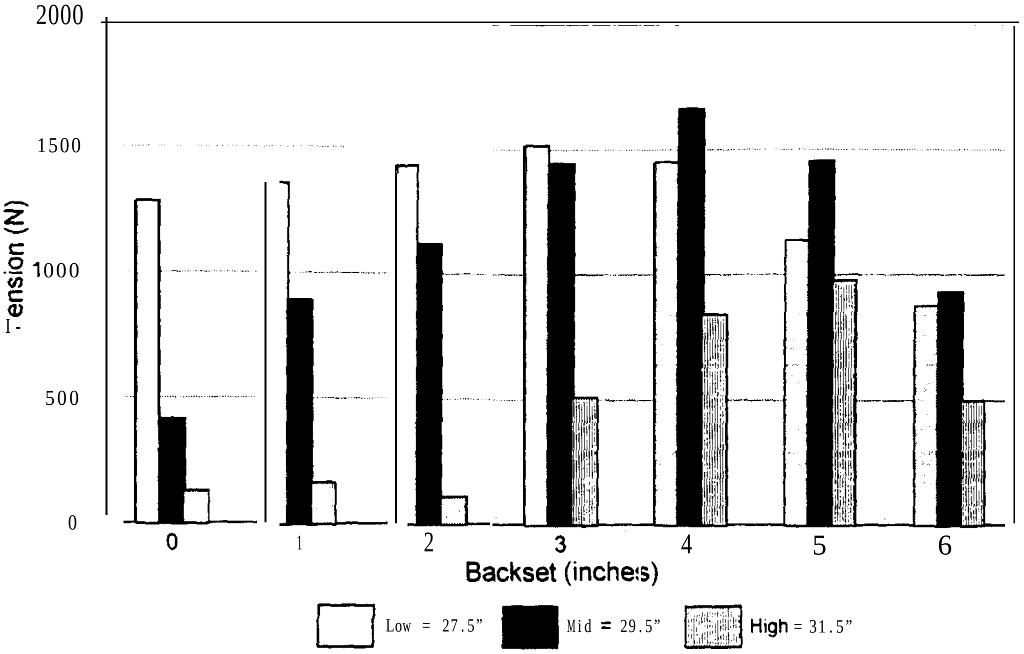


Figure 7: Maximum Tension at the Occipital Condyles. The most prominent effect of head restraint height was a decrease in tension as the height was increased. At larger backsets, the backwards rotation of the head when the head contacted the head restraint caused the tension to decrease.

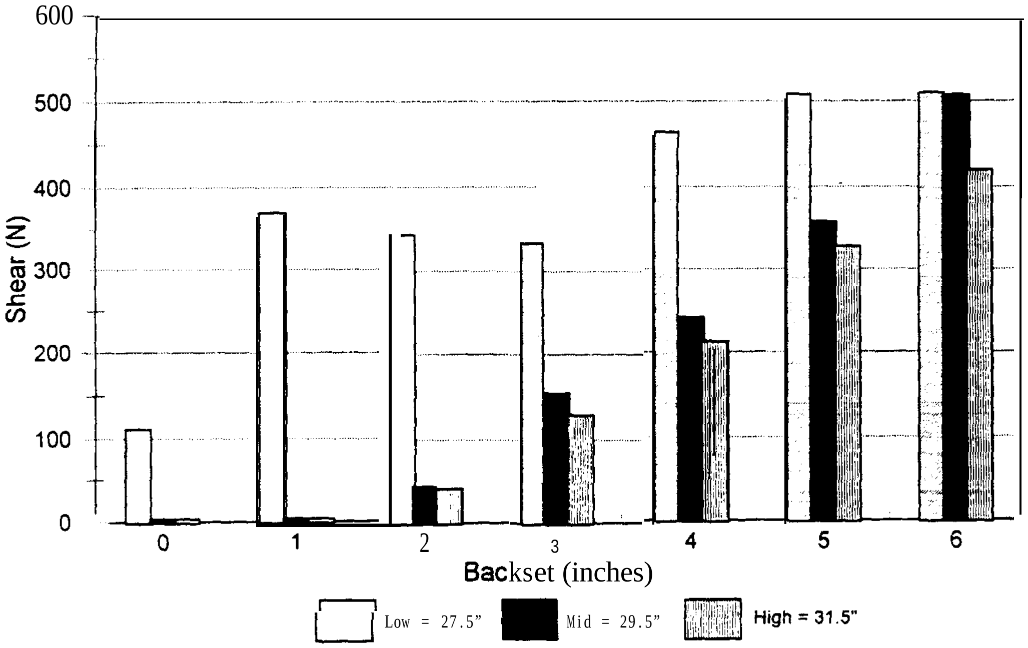


Figure 8: Maximum Positive Shear Force at the Occipital Condyles. The erratic trends in the shear forces at the low height are a result of the precise details of the rotation of the head over the head restraint and the location of the point of contact. At the intermediate and high heights, an increase in the backset caused delayed contact of the head with the head restraint and an increase in the peak shear forces.

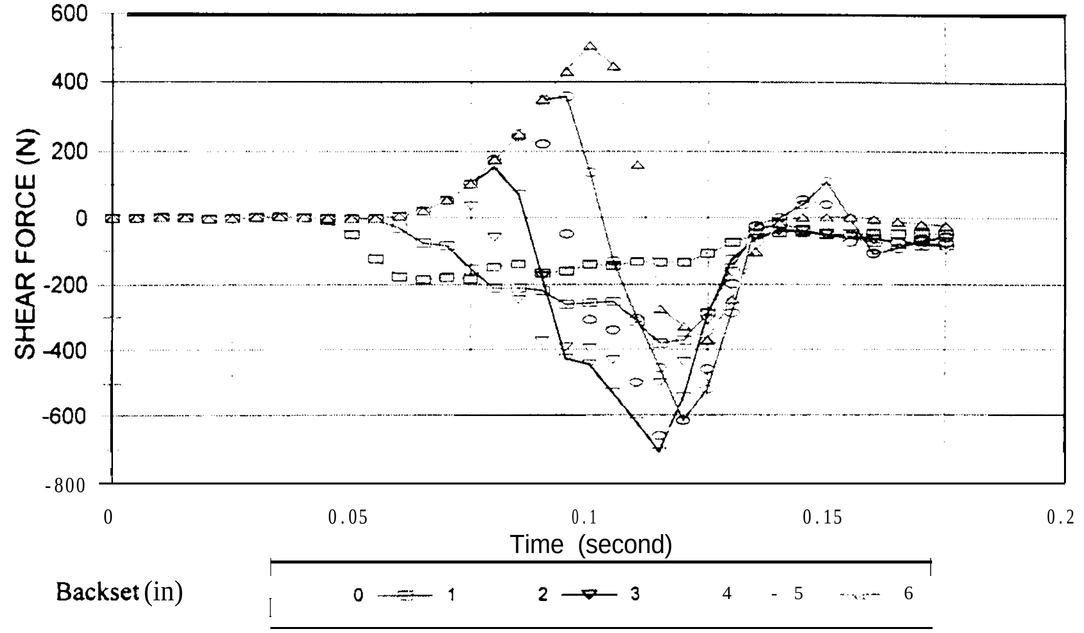


Figure 9: Shear Forces at the Occipital Condyles for the Intermediate Head Restraint Height of 29.5”. The peaks in shear forces correspond to the time of contact of the head with the head restraint. Thus, the time of contact and the peak shear force increase as the **backset** is increased.

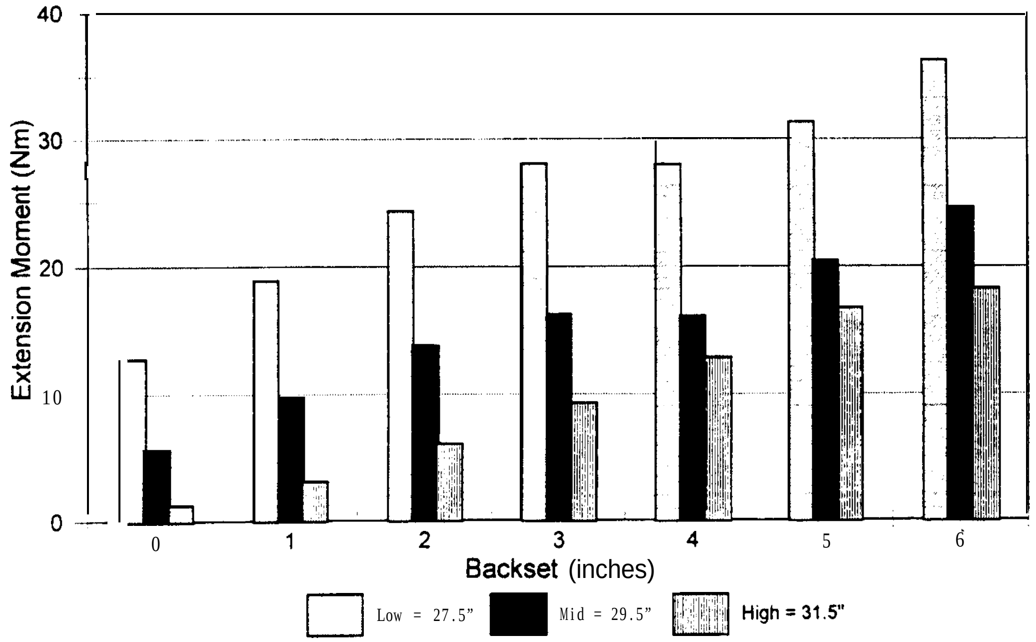


Figure 10: Maximum Extension Moments at the Occipital Condyles. Similar to the shear forces, the maximum extension moments decreased with increasing head restraint height and decreasing **backset.**

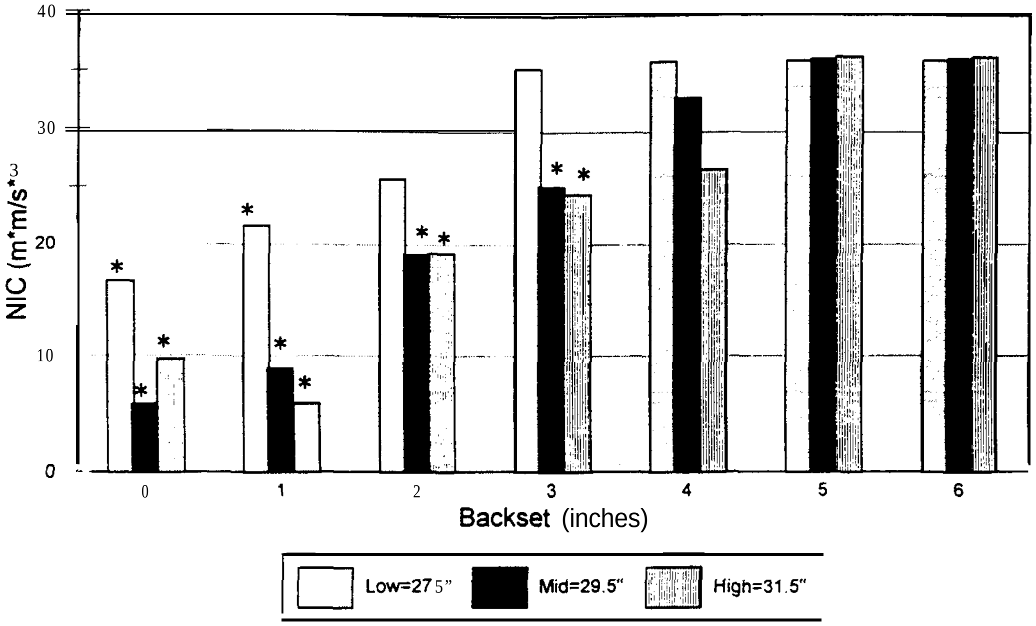


Figure 11: Maximum NIC value and NIC 50 values. The maximum NIC and NIC50 values were very similar, with the exception of cases where 50 millimeters of relative displacement were not reached or cases where the NIC50 value was negative. These cases are indicated in the figure with an asterisk (*). The proposed tolerance value for whiplash injury is 15 m^2/s^2.

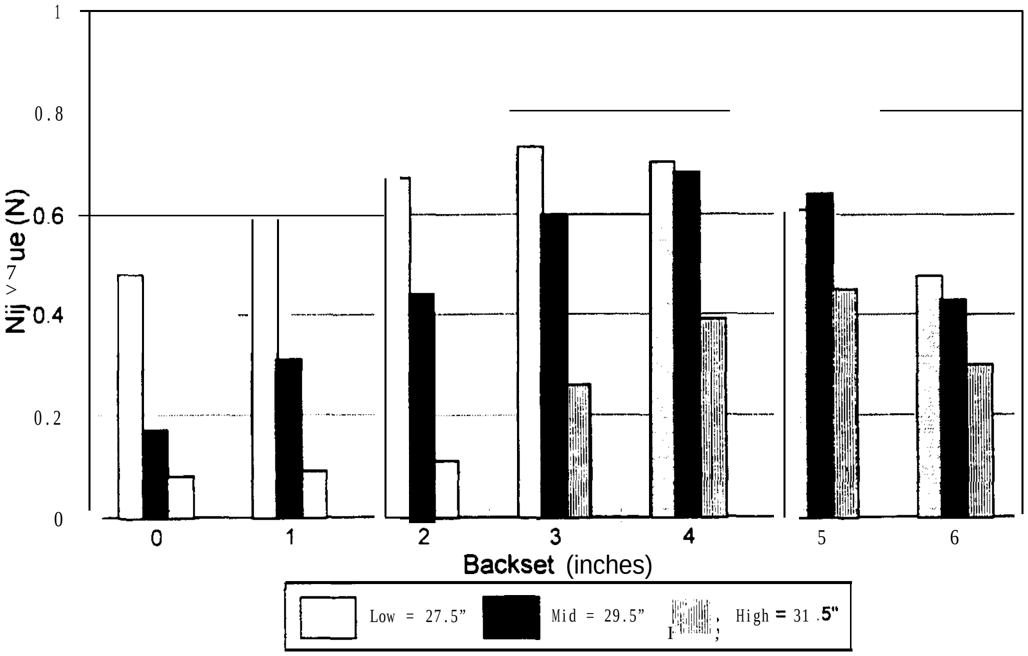


Figure 12: Maximum Tension-Extension Nij Value. The NTE values decreased with increasing head restraint height and decreasing backset. The NTE value were relatively low (<0.2) for the intermediate height at a backset of 0 inches and at the high height at a backset of 2 inches or less.