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Abstract

Human volunteer kinematic response to low speed rear-end collisions was investigated. Nominal 16 kph (10 mph) car-to-car impacts were conducted, using human volunteers and anthropomorphic dummies. The human volunteers were both male and female, aged 27 to 58 years, with various degrees of cervical and lumbar spinal degeneration (documented by MRI scan) at the time of the tests. Human volunteer response was monitored and analyzed via accelerometers and high speed film.

The impacts resulted in no injury to any of the human volunteers, and no objective changes in the condition of their cervical or lumbar spines. The results indicate a minimum injury tolerance to low speed rear-end impacts for males and females with various degrees of spinal degeneration. Kinematic responses of the head, mandible, upper torso and knees are discussed in light of existing theories regarding injury causation and tolerance.

Introduction

While extensive research has been conducted on occupant response to high speed vehicular impact, relatively little data regarding human occupant response to low speed impact exists. This trend has persisted in spite of the high costs to society from low speed vehicular impacts, particularly rear-end impacts. For purposes of this paper, low speed rear-end impact will refer to those impacts in which the struck vehicle sustains a change in velocity (Delta V) of 13 kph or less.

In order to explain and understand claimed injuries, researchers in both the medical and engineering disciplines have speculated about human tolerance to low speed rear-end impact, and hypothesized various injury mechanisms and occupant responses. However, this has been done largely without the benefit of useful experimental data.

The cornerstone of research into occupant response to high speed vehicular impact has involved the use of cadavers and anthropomorphic dummies (ATD's). The similarity of these surrogates to live humans in the high speed frontal and lateral impact environments has been reasonably well established in the medical and engineering literature over the past four decades. The logistic advantages of these surrogates for crash testing and research have led to several rear-end crash test studies using either cadavers or dummies [Severy et al. 1968, Clemens and Burrow 1972, Thompson 1990, Emori and Horiguchi 1990].

Occupant responses obtained from these rear-end studies employing surrogates remain of questionable significance, for several reasons. Cadavers clearly lack muscular input, which is likely an important contributor to occupant kinematics in a low speed rear-end impact. The "state-of-the-art" Hybrid III dummy is equipped with a relatively stiff neck, and an upper torso which was essentially designed on the basis of cadaver response to high speed frontal impacts. Most other anthropometric devices are similarly designed. As a result, existing mechanical surrogates may not be capable of accurately simulating live human response in low speed rear-end impacts. Recent research has quantified some important differences between humans and the Hybrid III in low speed rear-end impacts [Scott et al. 1993].

Computer simulations and mathematical models hold great potential for analysis of occupant response and injury mechanisms in vehicular crashes. However, those studies which have attempted to model occupant response to low speed rear-end impacts have done so without the benefit of an adequate database of useful experimental data for validation, and therefore remain of unknown significance.

Actual crash testing with human volunteers remains the only valid method to determine response and tolerance to low speed rear-end impacts.

Stapp [excerpted from Sances et al. 1986] conducted some of the earliest human tolerance tests between 1946 and 1951. Human volunteers were helmeted, and seated in an aeronautical seat mounted on a test sled equipped with a lap belt. The volunteers were exposed to rearward (positive x direction) accelerations of up to 35 g "without exceeding the limits of voluntary tolerance." While these tests are of questionable relevance to low speed automotive research, they give some insight into the tolerance of the human body to impact and acceleration.

The first documented low speed rear-end automobile crash tests using human volunteers were conducted by Severy et al. [1955]. He impacted stationary target vehicles with similarly massed bullet vehicles traveling at approximately 13 and 15 kph, with a lap-belted male volunteer seated in the target vehicle. The volunteer was described as relaxed for both impacts, although Severy indicated that the subject may have employed a "conditioned muscle reflex" during the second impact. Severy indicated that the crashes were "believed to be non-injury producing," and did not report any injury as a result of the impacts. The applicability of Severy's volunteer data to modern crashes may be compromised by the fact that the vehicles he used were not equipped with energy absorbing bumpers or head restraints, although arguably the addition of these features would only decrease the likelihood of injury. Severy's collisions are thus an initial indication of a minimum tolerance to injury in low speed rear-end collisions.

Mertz and Patrick [1967] conducted several sled tests which simulated rear-end impacts. In tests without head restraints, a tensed male volunteer was exposed to average decelerations of up to 3.2 g without injury. Exposures of up to 8.7 g average acceleration were tolerated by a tensed male volunteer in a rigid seat equipped with a flat, padded head restraint. The authors concluded that a "10 mph (16 kph) rear-end collision should be tolerable even in the case of the unsuspecting individual" and further that "it is physically possible for a person to withstand a 44 mile per hour (71 kph) rear-end collision with no injuries, provided his head is initially in contact with a flat headrest which is firmly attached to a rigid seat back."

McConnell et al. [1993] subjected four male volunteers, aged 45-56, to rear-end impacts using four contemporary vehicles. Stationary target vehicles with human occupants underwent Delta V's of approximately 3 to 8 kph. The bullet vehicles also contained human volunteers, and underwent Delta V's of approximately 3 to 9 kph. Three-point restraint systems were used in all cases, and at least one of the target vehicles was not equipped with head restraints.

In McConnell's test series, each of the subjects was exposed to between three and seven collisions over an eleven-day period. Two of the four subjects reported mild neck discomfort which lasted no more than five hours. One of the subjects experienced mild neck discomfort the morning following a day in which he was subjected to three collisions. This discomfort lasted three days. The authors concluded that collisions with a Delta V of 8 kph "appeared to be on the threshold for mild cervical strain injury for our repetitively exposed test subjects." The authors also stated that the "4 kph Delta V test runs were considered later by the participating physician test subjects to have been so very mild that a single exposure would have been unlikely to have resulted in any symptomatology."

West et al. [1993] subjected five male volunteers, aged 25-43, to multiple rear-end impacts in various contemporary vehicles. Seat back heights ranged from 71 to 81 cm. Peak target vehicle accelerations were 0.9 to 17.1 g, and the reported barrier equivalent velocities ranged from approximately 4 to 13 kph. Each volunteer was subjected to 10-12 rear-end impacts of increasing severity over a one day period.

The only symptoms reported in the West study were minor neck pains for two volunteers which lasted for one to two days. The authors indicated that these minor symptoms were likely the result of multiple impacts in a vehicle with poor head restraints [correspondence with West]. The authors concluded that "if proper head support is available, impacts with an equivalent barrier speed in excess of 8 kph can be tolerated without injury."

Siegmund and Williamson [1993] subjected two male volunteers to aligned bumper car collisions. A 25 year old volunteer was subjected to six rear-end collisions, while a 32 year old volunteer was subjected to three rear-end collisions, each over a two hour period. Target bumper car Delta V's ranged from approximately 6 to 8 kph, and the duration of the collisions (160 to 220 ms) were noted to be comparable to low speed automobile collisions.

The bumper car seat backs were approximately 47 cm high, and constructed of unpadded, rigid plastic. The volunteer exposed to six impacts attempted to relax his neck and shoulder musculature for each impact, while the other volunteer adopted an initial position with his head angled forward from vertical for each impact. Neither volunteer complained of any pain following the impacts; however, the authors noted that the volunteers were likely more physically fit than the general population, and that bumper car occupants may be better prepared to anticipate an impact than automobile occupants.

The majority of volunteer research conducted to date has considered essentially healthy male exposure to low speed rear-end and frontal impacts. No reference to pre-existing spinal conditions, or female exposure to rear-end impacts, was found. The current study endeavors to enhance the database of human tolerance to, and kinematics during, low speed rear-end impact, and to consider both the initial spinal condition of the volunteers, and female exposure to low speed rear-end impact.

A series of low speed rear-end impact tests was conducted in the present study, using instrumented male and female human volunteers, some of whom were subjected to pre and post collision MRI studies of the cervical and lumbar spines. Occupant kinematics are presented, and human tolerance to low speed rear-end impact is examined in light of this research.

Methodology

Six 16 kph (10 mph) car-to-car impact tests were conducted using 1981 and 1982 two-door hatchback Ford Escorts. The bullet vehicle was towed into the stationary target vehicle via an underground cable system. The impacts were bumper-to-bumper with no lateral or vertical offset. The tests were conducted over a two-day period in August, 1991.

Vehicles

Ford Escorts of model years 1981 and 1982 were chosen as both bullet and target vehicles for several reasons. Ford Escorts remained essentially unchanged during model years 1981 to 1985, with over two million vehicles built during that period [Ward 1984, Ward 1986]. They are highly represented on the road, and as such are likely involved in a substantial number of low speed rear-end collisions.

Both the front and rear bumpers of the Escorts are equipped with a piston-type energy absorbing bumper system. This bumper system is recognized as one of the most adept at withstanding low speed impacts without sustaining damage [IIHS 1993]. This characteristic permitted multiple tests on the same vehicle, without requiring vehicle body repair between tests.

The bumper assembly, consisting of the energy absorbing pistons and face bar, was replaced on both the bullet and target vehicles before each test, regardless of whether bumper damage was evident. Similarly, the front seats in the target vehicle were replaced before each test. This prevented any cumulative effects of loading on the bumpers or seat backs, enhancing the repeatability of the impacts.

Both doors were removed from the target vehicles to facilitate filming of the occupants. Ballast was added to compensate for the absence of doors to maintain original vehicle weight. Fastax rotary prism high speed cameras were mounted to the sides of the target vehicle, such that the occupants' entire bodies could be filmed.

Six crash tests were conducted. For the first five tests (Tests 1-5), the brakes on the target vehicle were automatically triggered to engage with full braking capacity at a time after impact, which was felt to simulate the braking response of a driver with a rapid perception-reaction time. The automatic application of brakes ensured repeatability between tests. The brakes on the target vehicle in Test 6 were not engaged at any time following impact. To avoid secondary impact, the bullet vehicle's brakes were triggered automatically after disengagement of the vehicle bumpers in all six tests.

Bullet and target vehicles were equipped with IC Sensors 200g uniaxial accelerometers mounted at the approximate vehicle static centers of gravity to measure longitudinal accelerations. Bullet vehicle impact velocity was recorded by a tape-switch-activated electronic speed trap.

The target Escorts contained seats which were equipped with the standard integrated head restraints. The seat back height was measured to be 72 cm, and the seat back geometry was reasoned to be typical of that in many current vehicles. The Escorts were equipped with an integrated three-point restraint system, with an inertial locking retractor. This was also reasoned to be typical of the majority of restraint systems currently on the road. Human occupants in both the bullet and target vehicles wore the standard restraint system in all tests.

Coordinate Systems

Several coordinate systems were used for data analysis. System 1 originated at the approximate center of gravity of the target vehicle, with the x axis parallel to the longitudinal axis of the vehicle (Vehicle Coordinate System). System 2 originated at the center of gravity of the occupant's head, with the x axis lying within both the Frankfort and saggital planes (Head Coordinate System). Systems 3 and 4 originated at the bases of the cervical and lumbar spines, respectively, with the x axes lying within the saggital plane, and perpendicular to the spine at the origin (Cervical and Lumbar Spine Coordinate Systems). System 5 originated at the top of the seat back, with the x axis lying perpendicular to the seat back (Seat back Coordinate System).

All systems used a right-hand rule, with the positive axes forward, to the right, and down for the x, y, and z axes, respectively (conforming with SAE J211). For clarity, the x axis in System 1 will be referred to as "X1", the y axis in System 2 will be referred to as "Y2", etc. The five coordinate systems are presented in Figure 1.

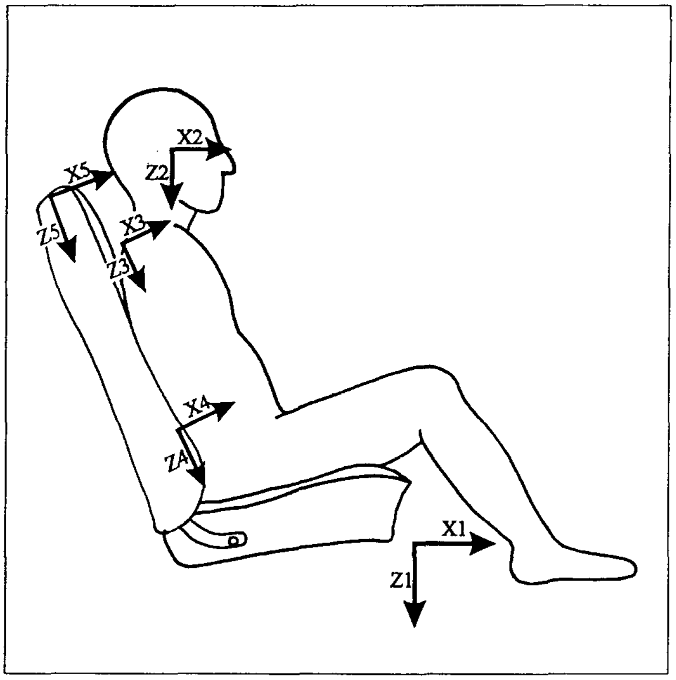

Figure 1: Coordinate Systems for Data Analysis

Occupants

Table 1 lists the sex, age, standing and seated heights, and weight for each volunteer.

Subject	Sex	Age	Standing Height (m)	Seated Height (m)	Weight (kg)
A	F	27	1.68	0.91	63.5
B	M	48	1.83	0.94	79.4
C	F	58	1.65	0.84	62.1
D	M	28	1.91	0.91	93.0
E	M	31	1.80	0.94	79.4

Table 1: Human Volunteer Descriptions

Both human volunteers and uninstrumented Hybrid II anthropometric dummies were used as front seat occupants in the target vehicles. In two of the tests an uninstrumented 50th percentile six-year-old child dummy (21.4 kg, seated height 0.65 m) occupied the rear seat of the target vehicle. Table 2 contains the seating arrangements for the target vehicle occupants in each test. Restraint usage is indicated in parentheses.

Day	Test	Driver	RF	RR
1	1	A (3 point)		
1	2	B (3 point)	Hybrid II (3 point)	
2	3	C (3 point)		6 yo ATD (none)
2	4	Hybrid II (3 point)	D (3 point)	
2	5	C (3 point)		6 yo ATD (lap)
2	6	A (3 point)	E (3 point)	

Table 2: Target Vehicle Seating Configurations

All volunteers were instructed to adopt a "normal" seating position, with the exception of the target vehicle driver in Test 5 (Subject C), who intentionally adopted an increased head-to-head restraint distance. Although all volunteers were aware of the purpose of the study, they were instructed to relax and simulate an unanticipated impact to the best of their abilities. Visual cues of the impending impact were defeated.

All volunteers were instrumented, with the exception of Subject E, who was filmed only. Triaxial blocks of IC Sensors 100g accelerometers were affixed at three locations on the volunteers' heads via a head band arrangement. The head band was lightweight so that the inertial characteristics of the occupant's heads were essentially unaffected. The headband was made of rubber, and fastened tightly to the occupant's head, such that movement of the band, relative to the head, was virtually impossible. The headband measured accelerations in the Head Coordinate System (X2,Y2,Z2) at the head static center of gravity, via an algorithm which considered the location of each triaxial block relative to anatomical landmarks which located the head static center of gravity. [Alem and Holstein 1977].

Uniaxial IC Sensors 100g accelerometers were affixed to each occupant's upper back at the base of the cervical spine, and lower back at the base of the lumbar spine, using skin adhesive. These accelerometers measured acceleration in the X3 and X4 directions, respectively.

Human subjects A and D were fitted with stiff, lightweight bite plates constructed by a dentist. The bite plates were affixed to the lower teeth, and had an external visual target attached so that any motion of the mandible could be captured on high speed film.

Data Acquisition

Excitation and amplification of the analog sensor data was accomplished by an on-board signal conditioning system. Data was transferred through a hard-wire bundle to a Kyowa RTP-650B analog recorder, recording at 10 kHz. The analog data was then transferred to a Metrabyte DAS-16F 10 Hz analog-to-digital converter.

The Fastax rotary prism high speed cameras were manually triggered and operated at a nominal film speed of 500 frames per second. High speed film images were transferred to video, and the video images were subsequently transferred to a computer through a Data Translations Quick Capture frame capture board. Analysis of the captured images was performed using Global Lab's image processing and analysis package. Each test was also videotaped from various angles so that additional qualitative analysis of occupant motion could be conducted.

Clinical Condition of Human Subjects

Human subjects A,B,C and D underwent examinations by a board certified orthopedic surgeon before the test series. These same test subjects underwent cervical and lumbar MRI's one month before, and three months after the crash tests. The pre and post test MRI's were randomly coded, and interpreted by two experienced radiologists in a double-blind study, to determine the presence of any objective changes in the condition of the occupants' cervical or lumbar spines.

Test Site

Tests were conducted at Mobility Systems and Equipment Company's Automotive Research Center in San Bernardino, California.

Results

Vehicle Response

The impact velocity of the bullet vehicles ranged from 15-16 kph. Target vehicle peak accelerations were 5-6 g, with pulse durations of approximately 100 ms, and a change in velocity for the target vehicle of approximately 8 kph. Time zero for both vehicle and occupant responses was established as initial bumper-to-bumper contact.

Szabo and Welcher [1992] provide a more detailed analysis of Ford Escort vehicle dynamics in 16 kph impacts.

General Occupant Response

A common overall occupant response was noted for the human occupants in the six tests. The occupant remained at rest, relative to the ground, as the vehicle was accelerated beneath him/her, resulting in rearward motion of the occupant relative to the vehicle interior. This rearward motion occurred during the initial 110-170 ms. Contact between the head and head restraint occurred in every test. The portion of the collision during which the occupant's upper torso moved rearward, relative to the vehicle, will be referred to as *Phase I* of the impact.

The rearward motion of the occupant was arrested by contact between the upper torso and seat back/head restraint. The occupant subsequently moved forward relative to the vehicle interior, until this motion was arrested by the restraint system. The portion of the collision during which the occupant's upper torso moved forward relative to the vehicle will be referred to as *Phase II* of the impact. The duration of Phase II ranged from 150-230 msec.

In five of the six tests, the restraint system retractor reeled back as the occupant's upper torso and seat back rotated rearward during Phase I, resulting in a "tightening" of the shoulder belt. The one exception occurred during Test 3, when the shoulder belt was inadvertently prevented from reeling back on Subject C by test hardware. In all tests, the inertial webbing retractors locked, and no contact between occupants and the vehicle interior occurred. Similar restraint system dynamics were reported by McConnell et al. [1993].

The translational and rotational motion of all occupants occurred exclusively in the saggital plane; no rotational movement occurred in the transverse or frontal planes for any occupant, and no significant lateral translational motion occurred.

The inertial restraint system locked in all tests, regardless of whether the brakes in the target vehicle were applied or not. Motions of the occupants in the unbraked vehicle were quite similar to those in the braked vehicles, indicating that a large component of the forward motion of the occupants within the vehicles was due to elastic behavior of the seat back, as opposed to deceleration of the vehicle.

The seat backs rotated rearward during Phase I of the impact. Seat back rotation was greater for heavier occupants, and in no case exceeded 10 degrees. Slight ramping up the seat back was experienced by all occupants, a finding also reported by McConnell et al. [1993].

None of the drivers maintained a secure grip on the steering wheel during Phase I, and in many cases their hands came off the steering wheel entirely. During Phase II the hands returned to, or remained in the vicinity of, the steering wheel, without re-establishing a firm grip.

The Hybrid II dummy motions were dissimilar to those of the human occupants. Decreased neck motion, different phasing of body segment motions, and an overall "stiff" response were observed for the dummies. The dummies' motions during Phase II of the collisions were especially dissimilar to those of the human volunteers. These differences were most likely attributable to the relatively stiff cervical and lumbar spines in the dummy, and dummy back anthropometry which is quite dissimilar to that of a human. The motions of the child dummies in the rear seat also appeared unrealistically stiff, although no human counterpart motion was available for comparison.

Given the observed differences between human and dummy responses, no further analysis of the dummy motions was conducted.

Clinical Condition of Human Subjects

None of the subjects complained of any orthopedic or neurologic symptoms, nor were abnormal findings noted at the time of the pre test orthopedic examination. Pre test MRI scans of the cervical and lumbar spines noted pathology in three of the four volunteers who underwent MRI's. The pre test MRI results are summarized in Table 3.

Subject	Cervical Spine		Lumbar Spine	
	Degree of Degeneration	Disk Bulge or Protrusion	Degree of Degeneration	Disk Bulge or Protrusion
A	1	1	2	0
B	0	0	2	1
C	1	2	0	0
D	0	0	0	0

0-normal
1-minor abnormality
2-moderate abnormality
3-severe abnormality

Table 3: Pre Test MRI Results
(adapted from Harter et al.)

Volunteers A,B,C, and E described a transient headache immediately post impact, which resolved spontaneously prior to exiting the target vehicle. Volunteer A, who underwent two rear-end impacts, reported transient, minor neck stiffness the morning following the first test. No other symptoms whatsoever were reported by any of the subjects in the one-year period following the tests.

No significant differences were found between the pre and post test MRI's, indicating no objective changes to the cervical or lumbar spines as a result of the impacts. A more complete discussion of the MRI analysis is contained in Harter et al. [submitted for publication].

Human Subject Kinematics

All occupant displacements were obtained from digitization of the high speed film, which was found to be accurate to approximately ±13 mm. Segment displacements are presented in coordinate systems deemed most meaningful for the measured parameter. Where appropriate, the response of Subject A was selected as a "representative response" for presentation.

Angular Displacements. Cervical spine extension/flexion is often cited as a potential injury mechanism in low speed rear-end impacts, and is often used synonymously with head rotation. It is important to note, however, that head rotation alone is not necessarily representative of cervical spine extension/flexion. Cervical spine extension/flexion is properly defined as rotation of the head *relative to the upper torso* [Luttgens and Wells 1989, Tortora 1989]. Since both the head and upper torso undergo rotation during a rear-end collision, a measurement of head rotation alone in a rear-end collision is meaningless with respect to cervical spine injury mechanics.

Head-to-torso rotation in the current study is thus defined as rotation of the Z2 axis (head) relative to the Z3 axis (upper torso).

Since each occupant adopted a somewhat unique initial position with respect to the head-to-torso angle, certain normalizing definitions were established.

Anatomical position was defined as the standing anatomical position. Kendall and McCreary [1983] defined the anatomical position as "erect posture, face forward, arm at sides, palms of hands forward with fingers and thumbs in extension." Kendall and McCreary designated the anatomical position as the zero position for defining and measuring joint motions.

In the anatomical position, the head is oriented such that the external acoustic meatus and the inferior orbital rim form a horizontal plane, which is also referred to as the Frankfort plane [Snyder et al. 1975].

Seated position was defined as the position adopted by each occupant while in a comfortable seating position within the vehicle. Due to the angle of the seat back, the torso is inclined from vertical while seated in the vehicle. Since each subject had a tendency to maintain the Frankfort plane essentially parallel with the ground, the subjects' heads were rotated forward with respect to the torso in the seated position, as compared to the anatomical position.

The head-to-torso angle relative to the anatomical position was deemed the most meaningful measurement in terms of cervical injury potential, as opposed to that relative to the seated position. All head-to-torso rotation references in this study are thus relative to the anatomical position, and not to the seated position.

The sign convention dictated that rearward (nose up) head-to-torso rotation be positive, while forward (nose down) head-to-torso rotation be negative. Cervical flexion is thus negative head-to-torso rotation, while cervical extension is positive.

Figure 2 shows the head-to-torso rotation for all subjects. The initial cervical flexion angle of the subjects ranged from 18-30 degrees in the seated position.

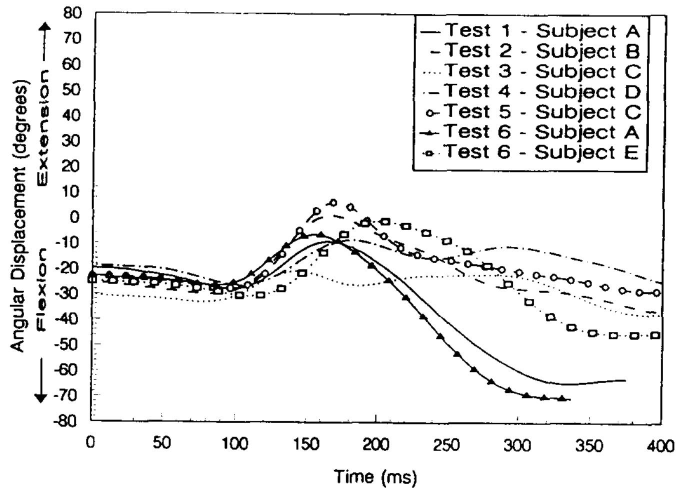

Figure 2: Head-To-Torso Rotation

Within the first 100-120 ms, all subjects exhibited slight flexion of the head relative to the torso (less than 10 degrees). This can be explained by the fact that as the upper torso rotated rearward during the initial moments of the impact, the head tended to remain in its initial position, resulting in the head rotating forward slightly relative to the torso. When the seat back comes in contact with the upper torso, the upper torso begins to accelerate forward relative to the ground, and the expected rearward rotation of the head relative to the torso during Phase I begins.

Only two subjects actually moved into a position of cervical spine extension relative to the anatomical position. However, in both cases, the extension was limited to less than 10 degrees. This was well within the reported normal voluntary cervical extension limit of approximately 75 degrees [Buck et al. 1959, Kottke and Mundale 1959].

No chin-to-chest contact occurred for any subject in the test series. Subject A achieved approximately 70 degrees of cervical flexion during Phase II of two impacts. This is the reported normal voluntary limit [Buck et al. 1959, Kottke and Mundale 1959], but was well within Subject A's actual voluntary flexion limit of approximately 90 degrees. No other subject's cervical spine flexion exceeded 45 degrees.

Each subject exhibited similar patterns of head-to-torso rotation during Phase I of the impact, regardless of sex, age, or height.

All subjects exhibited relatively similar patterns of head-to-torso rotation during Phase II of the impacts, with the exception of Subject C in Test 3 and Subject D in Test 4. In Test 3, the belt was inadvertently prevented from recoiling during Phase I of the impact, resulting in a second peak in cervical flexion upon delayed upper torso contact with the locked shoulder belt at about 180 msec. Review of the high-speed film for Test 4 revealed that Subject D may have anticipated the impact somewhat, possibly resulting in a decreased premotor reaction time, and therefore a quicker initiation of the upper body musculature in response to upper torso motion. This characteristic of response to stimulus is described by Schmidt [1988], and may be responsible for the double-peaked characteristic of cervical flexion during Phase II of Test 4 for Subject D.

No relative motion was observed between the mandible biteplates and the heads of Subjects A and D, indicating no motion of the jaw relative to the cranium throughout the collisions for either subject. Those occupants not equipped with biteplates also did not appear to experience mandible motion relative to the cranium in the other tests.

Linear Displacements. For each test, displacement of the head, shoulder, wrist, and knee were obtained. The positions of Subject A's head, shoulder, wrist, and knee relative to the vehicle in Test 1 are shown in Figure 3.

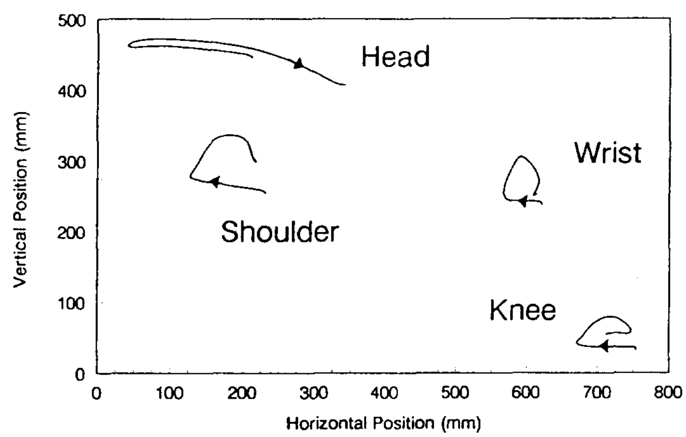

Figure 3: Head, Shoulder, Wrist and Knee Position in Vehicle Coordinate System (Subject A-Test 1)

Maximum forward and rearward displacements of the head, shoulder, knee and wrist, relative to the vehicle (X1 axis) for all occupants are listed in Table 4. All displacements are relative to the initial position of each segment. "Rearward" displacements represent the maximum linear X1 displacement of each segment, aft of its initial position, during Phase I of the impact; "forward" displacements represent the maximum linear X1 displacement of each segment, forward of its initial position, during Phase II of the impact. A negative forward displacement indicates that the segment did not return to its initial position during Phase II of the impact.

Test	Subject	Head		Shoulder		Wrist		Knee	
		rwd	fwd	rwd	fwd	rwd	fwd	rwd	fwd
1	A	170	130	104	-14	55	-5	83	-6
2	B	211	203	121	96	115	74	104	-15
3	C	179	268	112	197	117	68	73	15
4	D	209	153	176	118	238	-41	76	5
5	C	244	150	137	117	87	32	67	15
6	A	176	150	118	19	81	0	87	46
6	E	266	111	153	-3	159	75	103	4

All values are in millimeters.

Table 4: Maximum Head, Shoulder, Wrist and Knee X1 Displacements

Initial head-to-head-restraint distance for occupants A,B,C(Test 3),D and E ranged from 71 to 114 mm. The initial distance for occupant C(Test 5), who intentionally adopted an exaggerated head-to-head restraint offset, was 132 mm. The head dynamic displacement, relative to the seat back (X5 axis), for all subjects is presented in Figure 4.

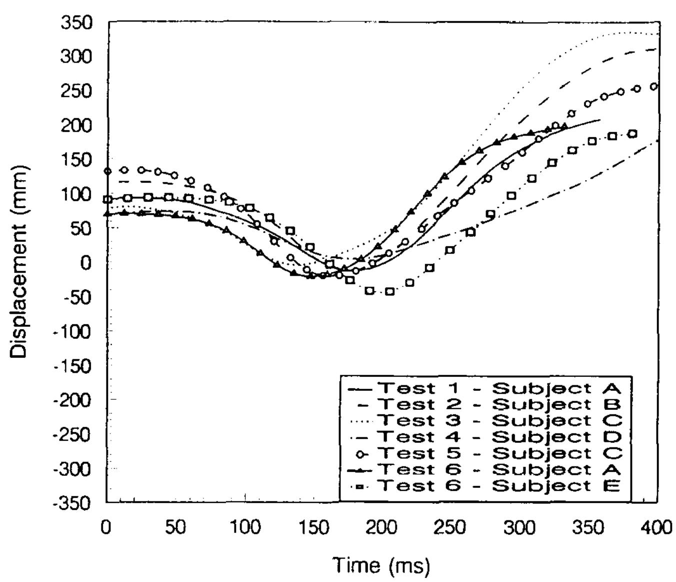

Figure 4: Head X5 Displacement

Figures 5 and 6 show the shoulder and knee displacements relative to the vehicle (along the X1 axis), respectively. All initial positions were normalized to zero for graphical comparison.

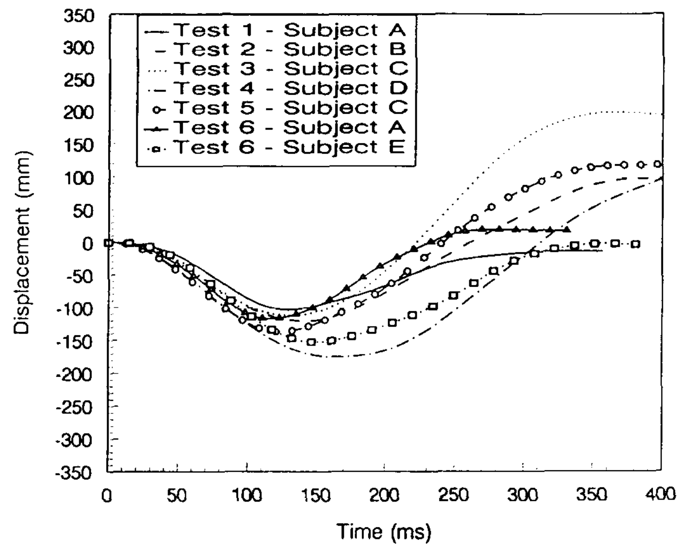

Figure 5: Shoulder X1 Displacement

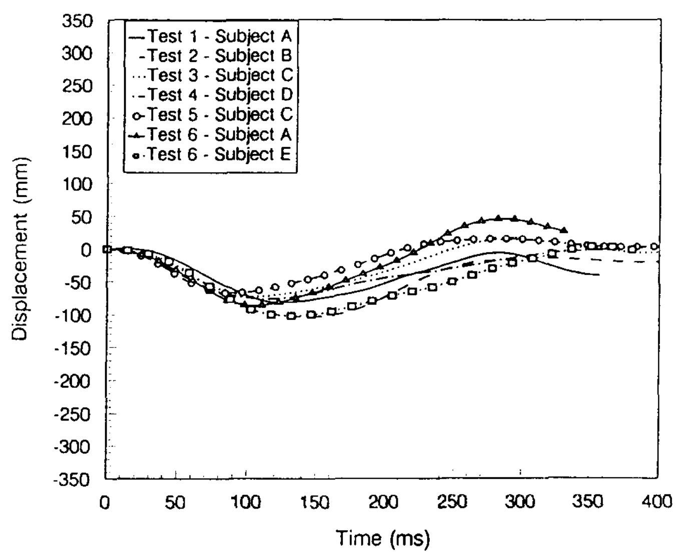

Figure 6: Knee X1 Displacement

Regardless of each occupant's initial head-to-head restraint distance, age, gender, or seated height, no significant variation in head, shoulder, or knee linear displacement was demonstrated during Phase I. As the event progressed, body motions of these segments remained essentially similar for all occupants; however, individual variations became more apparent during Phase II.

In most cases, during rebound, the occupant's head and shoulder continued forward past their initial positions relative to the vehicle. Neglecting the test in which the shoulder belt's retraction was interfered with (Subject C-Test 3), the maximum forward head excursion was less than 205 mm from the head's initial position, and the maximum forward shoulder excursion was less than 120 mm from the initial shoulder position. In some cases, the shoulder did not move forward past its initial position relative to the vehicle.

The knees of the subjects did not move significantly forward from their initial positions, relative to the vehicle, for all occupants (15 mm or less), with the exception of Subject A in Test 6. The observed 46 mm forward motion of the knee was a result of the subject extending her knee and hip during this test, and not associated with motion of the entire lower torso.

Velocities. The maximum rearward shoulder velocity, relative to the seat back, was less than 1.0 m/s for all occupants, which is less than the vehicle's velocity change. Occupant shoulder contact with the seat back occurred at less than the Delta V of the vehicle largely because of the rearward motion of the seat back during Phase I. The maximum forward shoulder velocity relative to the seat back was less than 2.1 m/s.

Knee velocity relative to the vehicle for all occupants was less than 1.5 m/s throughout each impact.

Shoulder velocity relative to the seat back, and knee velocity relative to the vehicle, for Subject A in Test 1 are shown in Figures 7 and 8, respectively. Positive velocity indicates forward motion, while negative velocity indicates rearward motion, relative to the vehicle.

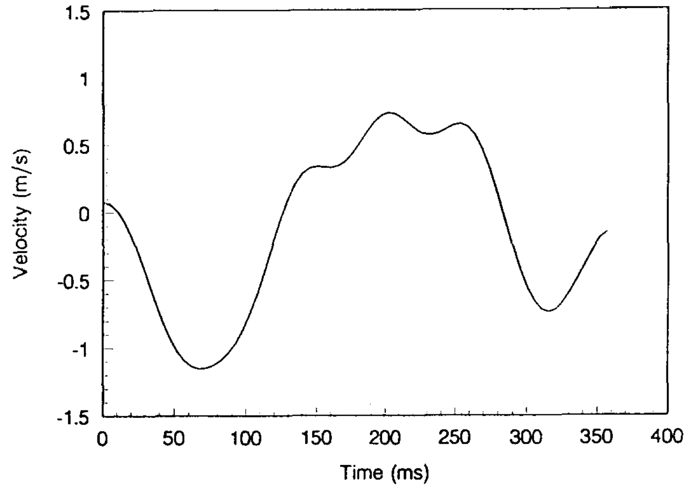

Figure 8: Knee Velocity Relative to the Vehicle (Subject A-Test 1)

Accelerations. Figures 9 and 10 show the head X2 and resultant accelerations for Subject A in Test 6, respectively.

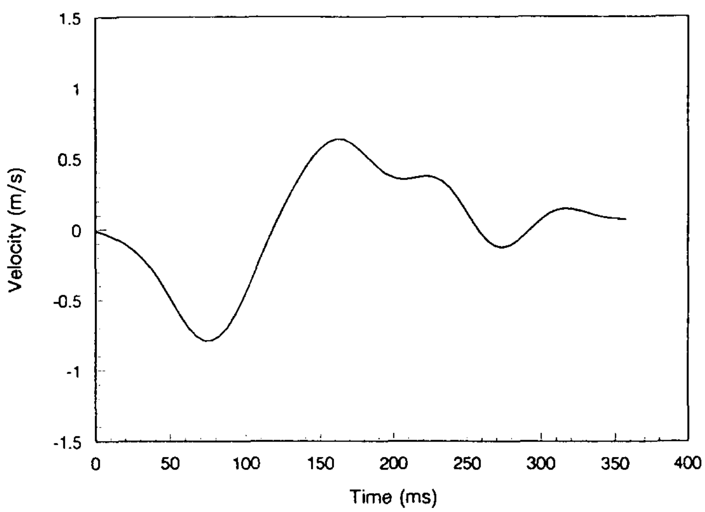

Figure 7: Shoulder Velocity Relative to the Seat Back (Subject A-Test 1)

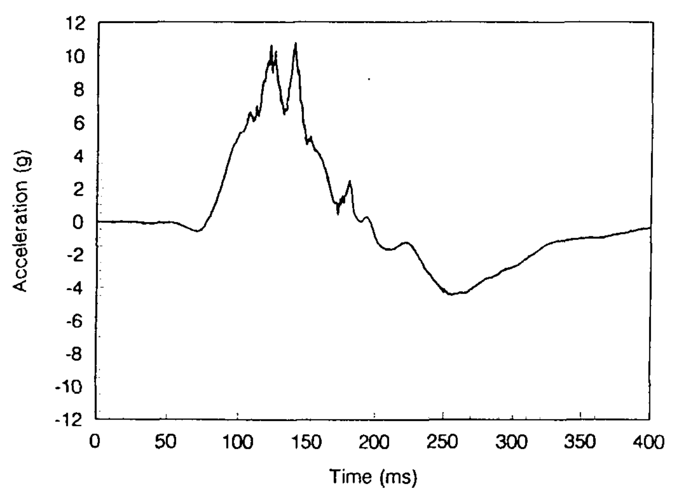

Figure 9: Head X2 Acceleration (Subject A-Test 6)

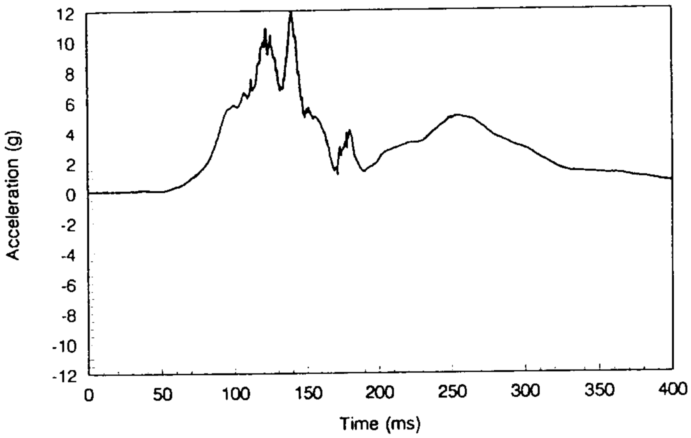

Figure 10: Head Resultant Acceleration (Subject A-Test 6)

Head accelerations were primarily translational, primarily forward in direction, and occurred almost exclusively in the sagittal plane. Tangential acceleration at the periphery of the skull was not found to differ significantly from center of mass acceleration, and as a result, it is concluded that the angular head accelerations were relatively low. Resultant peak head accelerations ranged from 10.1 to 13.7 g, and associated local pulse durations ranged from approximately 15 to 40 msec.

Figures 11 and 12 show the X3 acceleration of the lower cervical spine, and the X4 acceleration of the lumbar spine for Subject A in Test 6, respectively.

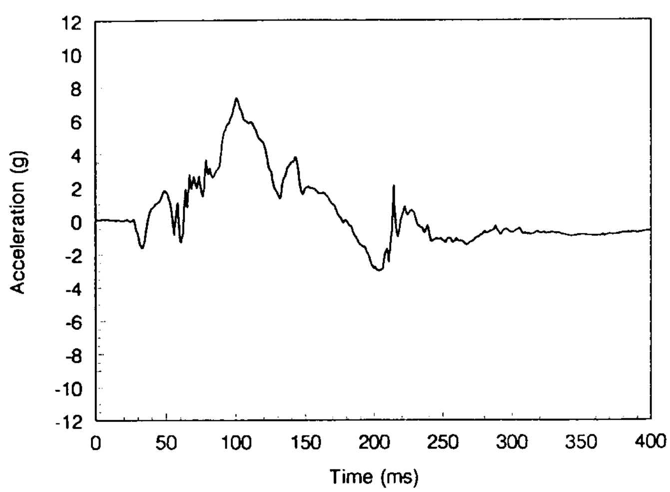

Figure 11: Lower Cervical Spine X3 Acceleration (Subject A-Test 6)

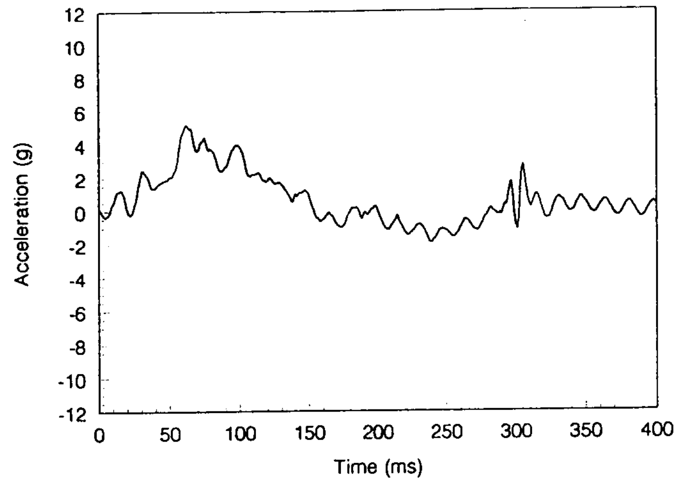

Figure 12: Lumbar Spine X4 Acceleration (Subject A-Test 6)

Maximum peak cervical X3 accelerations ranged from 4.5 to 7.4 g. Lumbar spine data was obtained for Tests 3 and 6 only, due to failed channels. Maximum peak lumbar X4 accelerations for these two tests were 3.9 g and 5.2 g.

Table 5 contains maximum and minimum peak accelerations for the head, cervical spine and lumbar spine for all tests.

Parameter	Coord. System	Test	Subject	Maximum (g)	Minimum (g)
Head Resultant Acceleration	2	1	A	11.8	0.0
		2	B	Failed channel	
		3	C	13.7	0.0
		4	D	10.1	0.0
		5	C	11.4	0.0
		6	A	11.9	0.0
Cervical Spine Acceleration	3	1	A	6.6	-2.0
		2	B	Failed channel	
		3	C	5.6	-4.1
		4	D	4.5	-2.8
		5	C	6.5	-2.7
		6	A	7.4	-3.0
Lumbar Spine Acceleration	4	1	A	Failed channel	
		2	B	Failed channel	
		3	C	3.9	-3.1
		4	D	Failed channel	
		5	C	Failed channel	
		6	A	5.2	-1.9

Table 5: Head, Cervical Spine and Lumbar Spine Acceleration Peaks

Acceleration-time histories for all tests showed that the lumbar spine is accelerated first, followed by the cervical spine, then the head, as would be expected. A phenomenon seen in all tests was initial slight rearward acceleration of the head. This initial rearward acceleration occurs because the lower torso is accelerated forward before the upper torso. As this occurs, the center of pressure is below the center of mass of the thorax and head, causing points above the center of mass to be accelerated slightly rearward, initially. This initial small rearward acceleration is of little consequence to the ensuing dynamic response of the head and upper torso.

Discussion

Both females and subjects with pre-existing spinal pathology were subjected to "unanticipated" rear-end collisions with a Delta V of 8 kph. Four of five volunteers described a transient headache immediately post impact, which resolved spontaneously prior to exiting the vehicle. One female volunteer reported minor, transient neck stiffness the morning following a test. No other symptoms were reported in the one year period following testing. No objective changes in spinal condition were found pre and post impact, as verified by MRI studies.

The present study enhances the existing database of volunteer studies which support the premise that, for restrained occupants with a head restraint available, single exposure to a rear-end collision with a Delta V of 8 kph or less is within human tolerance levels, and extends the database to include females and those with some degree of pre-existing spinal pathology.

In spite of the fact that human volunteers in the present study differed in sex, age, height, weight and initial spinal condition, kinematics for all occupants were similar, especially during Phase I of each impact, as the occupant moved rearward relative to the vehicle. During the subsequent rebound (Phase II), occupant responses began to diverge somewhat. This was most likely due to variations in muscle recruitment patterns during the latter stages of the event.

No occupant underwent cervical spine hyperextension or hyperflexion in the tests. This finding is consistent with that of McConnell et al. [1993] and West et al. [1993], who reported that cervical spine extension and flexion for their human subjects in rear-end collisions of similar severity were maintained within voluntary physiological limits. In 5 of 7 cases in the present study the occupants did not achieve a position of anatomical cervical extension, and remained in anatomical flexion throughout the event. The Ford Escort seat back and head restraint thus functioned not only to prevent significant cervical extension for different human subjects, but, in many cases, to prevent cervical extension entirely.

The restraint system consistently prevented contact between the occupants and forward structures within the vehicle interior during rebound. In all cases, the occupant's knee was not moved more than 15 mm forward of its initial position. Moreover, forward knee velocity, relative to the vehicle, remained less than 1.5 m/s, which is less than reported normal walking speed [Thompson 1991]. It appears that the potential for knee contact or knee injury for restrained occupants in rear-end collisions with a Delta V of 8 kph or less is extremely low.

Foreman and Croft [1988] and Neel et al. [1988] hypothesized that the upper torso rotates about the shoulder harness during rebound (Phase II), creating torsional stress on the lumbar and thoracic spines, especially if some slack is present in the shoulder harness. No such motion was observed in the present study, even for the case in which belt slack was inadvertently introduced. Motion occurred exclusively in the saggital plane, with no perceptible rotation in the transverse plane. Negligible relative motion was observed between the thoracic and lumbar spines of the human subjects in the present study. The likelihood of significant low back injury for a restrained occupant in a rear-end impact with a Delta V of 8 kph or less is considered remote, a conclusion also reported by West et al. [1993].

Head accelerations were primarily the result of contact with the head restraint, and were well below established human tolerances for head injury. A Holloman Air Force Base human volunteer head impact test produced a peak head acceleration of 75 g, over a 70 ms pulse duration, without report of concussion or other indications of brain injury [Hodgson and Thomas 1972]. Sled impact tests demonstrating voluntary human tolerance to head accelerations in excess of 50 g, with pulse durations up to 50 ms, without report of concussion or brain injury, are referenced and discussed by Gadd [1966] and Versace [1971]. Furthermore, the Wayne State Tolerance Curve average acceleration tolerance value corresponding to a 15-40 msec pulse duration is approximately 50 g [Fan 1971]. By contrast, volunteers in the present study sustained peak head accelerations of less than 14 g, for durations of less than 40 msec. Impact-induced head injury would thus seem extremely unlikely for a restrained occupant in a rear-end collision with a Delta V of 8 kph or less.

An often quoted parameter in low speed rear-end research is the so-called "head acceleration multiplication factor." This is simply the ratio of the peak head acceleration to the peak vehicle acceleration, and is sometimes implied as an indication of cervical injury potential when the ratio exceeds 2. Given that the majority of the head acceleration is the result of impact with the head restraint and is a function of head restraint stiffness, and not a reflection of internal forces generated within the neck, any attempt to relate such a parameter with cervical injury potential is futile. Indeed, the "head acceleration multiplication factors" for the present study ranged from about 2 to 3, without any ill effects.

Several studies have proposed a mechanism for temporomandibular joint injury in rear-end collisions [Weinberg and Lapointe 1987, Schneider et al. 1989]. These studies hypothesized that the jaw opens during Phase I of the impact, as the head rotates rearward, resulting in undue stress imposed on the temporomandibular joints. The present study does not support such a mechanism, as no jaw motion relative to the cranium was observed for any human subject during the rear-end impacts. This finding is in agreement with West et al. [1993], who also found no jaw motion during rear-end collisions, and supports the logical arguments proposed by Howard et al. [1991] and Orner [1992].

The anthropomorphic dummies used in this study were inappropriate human surrogates for low speed rear-end crash testing, supporting the conclusions of Scott et al. [1993]. Deficiencies in the dummy used include improper back anthropometry, and cervical and lumbar spines that are stiffer than those of the "unaware" human. Given the complexities of human volunteer research, a biofidelic anthropomorphic dummy for use in low speed rear-end crash tests would be of benefit.

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