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Aircrew Ejection Injury Analysis and Trauma Assessment Criteria

Bruce M. Bowman

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AIRCREW EJECTION INJURY ANALYSIS
AND TRAUMA ASSESSMENT CRITERIA

Final Report

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16. Abstract

A comprehensive literature review was conducted to document the types of injury that can occur during ejection in emergency escape from fighter and attack aircraft. On the basis of the literature an injury priority analysis was performed and criteria for trauma assessment were documented. The results of this research are pertinent to the application of an ejection test manikin in Navy studies of automated escape systems. The scope of the study was limited to the phases of the escape sequence that precede complete egress from the aircraft.

In the Injury Priority Analysis subtask the most important observational ejection-related injuries are identified. This establishes the types of injury that are most important to study and, therefore, the types of dynamic response data that

- (continued on next page) -

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16. Abstract (continued)

should be determined in manikin ejection tests. The Trauma Assessment Criteria subtask establishes how the dynamic response measures determined from testing should be interpreted in terms of injury potential.

Injury Priority Analysis -- It is clear from even a cursory review of the literature on ejection-related injuries that spinal column fractures are the dominant and most severe injuries that result during ejection-seat acceleration. The *central findings* of the current study as regards injury types and rates are

1) Fractures of lower thoracic and upper lumbar vertebrae, from T11 to L2, are the dominant major injuries that occur prior to complete egress from the aircraft during aircrew ejection. Such fractures occur in typically 20 percent of all ejections (7 to 47 percent, depending on the data base examined). Thoraco-lumbar fractures are most common at T12 and L1.

2) Fractures of the cervical vertebrae are five to seven times less common than fractures in the thoraco-lumbar spinal column, but they are nonetheless important since they are sometimes fatal and are much more often associated with permanent, major disability. Cervical fractures are most common at C2, C5, and C6.

3) Fatal head and neck injuries of nonspecific type may occur with significantly higher rates (although still small) for through-the-canopy systems without fragmentation devices than for jettisoned-canopy systems.

Trauma Assessment Criteria -- Methods were documented for relating dynamic response parameters that can be measured with manikins under experimental conditions to injuries that may be sustained by an aircrew member in a real-world ejection--specifically, the types of injuries identified in the Injury Priority Analysis subtask.

It is certain that a manikin neck that is too simple will be incapable of predicting all of the types of failure that can occur in a human neck, and it is of particular importance that the range of validity of the manikin neck be established by comparison of results from tests with manikins and cadavers or, indirectly, by confirmation of proper manikin prediction of ejection-related injuries seen in operational conditions.

Various useful criteria are available, and described, for prediction of thoraco-lumbar spinal fractures. For various reasons adoption of conservative neck-injury criteria for trauma assessment in ejection studies is recommended.

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EXECUTIVE SUMMARY

The research reported here was conducted by the University of Michigan Research Institute (UMTRI) as a subcontractor to Conrad Technologies, Inc. This study, *Aircrew Ejection Injury Analysis and Trauma Assessment Criteria*, is one task of a larger research effort conducted by Conrad Technologies, Inc., for the Naval Air Warfare Center (NAWC) under Contract No. N62269-91-C-0225.

The overall goals of the Conrad Technologies research for NAWC relate to several aspects of increasing the safety of aircrew of Navy aircraft, both in normal operation of the aircraft and in emergency situations when it may be necessary to abandon the aircraft.

A comprehensive literature review was conducted to document the types of injury that can occur during ejection in emergency escape from fighter and attack aircraft. On the basis of the literature an injury priority analysis was performed and criteria for trauma assessment were documented. The results of this research are pertinent to the application of an ejection test manikin in Navy studies of automated escape systems.

The scope of the study was limited to the phases of the escape sequence that precede complete egress from the aircraft; i.e.,

- aircraft maneuvering
- pre-escape positioning of crewmember
- ejection boost
- helmet impact with the canopy
- exposure to a rocket exhaust

Navy and Air Force researchers have reported that in 62 to 84 percent of major-injury cases ejection forces were judged responsible for primary injuries. Typical corresponding numbers reported for windblast and parachute opening shock injuries are 28 and 10 percent. Injuries that can occur post-egress were not studied in this research.

In the Injury Priority Analysis subtask the most important observational ejection-related injuries are identified. This establishes the types of injury that are most important to study and, therefore, the types of dynamic response data that should be determined in manikin ejection tests. The Trauma Assessment Criteria subtask establishes how the dynamic response measures determined from testing should be interpreted in terms of injury potential.

Injury Priority Analysis -- Despite the certain presence of some amount of error in existing ejection data bases, as well as nonspecificities, which can make interpretations uncertain, it is clear from even a cursory review of the literature on ejection-related injuries that spinal column fractures are the dominant and most severe injuries that result during ejection-seat acceleration. Regardless of the region of the spinal column considered--thoracic, lumbar, or cervical--the occurring fracture injuries are predominantly anterior-lip crush fractures that result from flexion-compression loading. The *central findings* of the current study as regards injury types and rates are

1) Fractures of lower thoracic and upper lumbar vertebrae, from T11 to L2, are the dominant major injuries that occur prior to complete egress from the aircraft during aircrew ejection. Such fractures occur in typically 20 percent of all ejections (7 to 47 percent, depending on the data base examined). Thoraco-lumbar fractures are most common at T12 and L1.

2) Fractures of the cervical vertebrae are five to seven times less common than fractures in the thoraco-lumbar spinal column, but they are nonetheless important since they are sometimes fatal and are much more often associated with permanent, major disability. Cervical fractures are most common at C2, C5, and C6.

3) Fatal head and neck injuries of nonspecific type may occur with significantly higher rates (although still small) for through-the-canopy systems without fragmentation devices than for jettisoned-canopy systems.

There is virtually no disagreement among the authors of the reviewed references regarding the most important injury types seen in ejection data bases, although it is noted here that the third finding above is based on fewer relevant references.

Numerous parameters were considered for their possible importance in influencing injury rates. They were type of aircraft maneuver, crewmember pre-escape positioning, ejection boost forces, helmet impact with the canopy, aircraft speed, severity of maneuvers, mission requirements, and crewmember physiology and anthropometry. Of the parameters that have bearing on injuries that occur before complete egress, ejection boost forces and crewmember pre-escape positioning were found to be of greatest importance. Helmet impact with the canopy may also be important. Crewmember physiology and anthropometry were not important factors. Regarding crewmember pre-escape positioning it is of prime importance for the reduction of vertebral fracture rates for crewmembers to be seated erectly, with buttocks, shoulders, and head back; the torso restraint should be tight, but not so tight as to force the shoulders down.

Trauma Assessment Criteria -- Methods were documented for relating dynamic response parameters that can be measured with manikins under experimental conditions to injuries that may be sustained by an aircrew member in a real-world ejection--specifically, the types of injuries identified in the Injury Priority Analysis subtask, viz., fracture injuries of the thoraco-lumbar and cervical regions of the spinal column.

Of the techniques that can be used to relate manikin dynamic responses to the potential for thoraco-lumbar spinal fracture, two types are most useful: 1) measurement of whole-body response and 2) measurement of compression loadings along the "spine" of the manikin. For the whole-body response technique, two different models are used, viz., the Dynamic Response Index (DRI) Method and the Acceleration Exposure Limit Method. They calculate the DRI and the Injury Risk Criterion, respectively. The models are similar, but the DRI is based on Z-axis response only while the Injury Risk Criterion is based on independent X-, Y-, and Z-axis responses. For either of these whole-body response methods to be useful in a manikin study, the manikin must have Z-axis spinal impedances that are similar to those of a human ejectee. Both methods are calibrated for injury-probability prediction on the basis of observational injury rates and cadaver tests. The technique that measures compression loadings along the spine of the manikin requires injury criteria for compression fracture of human thoraco-lumbar vertebrae. Such data are available and are documented in the report, together with means for predicting injury probabilities for associated maximum loads.

The only technique that appears capable of relating manikin dynamic responses to the potential for cervical spine fractures is direct measurement of manikin neck loads--both forces and moments--for comparison with cadaver neck injury data, which are given in this report. The mechanisms for vertebral fracture in the neck, however, are complex, depending not only on the loads on and ultimate strengths of the vertebrae but also very sensitively on initial positions and the conditions of loading. It has been found that nonalignment axially of the head, neck, and torso can reduce by half the neck compressive loads necessary to cause cervical fracture. Various authors find that peak impact force in S-I head impacts and peak compressive neck loads in quasistatic loading are not good predictors of cervical injury. One study finds that in S-I impact tests peak head linear velocity is the best indicator of injury of all response parameters measured. Of the impact parameters examined in that study, the integral of the impact force-time curve (the impact impulse) was the most consistent indicator of cervical injury.

Low values for compressive failure strengths of cervical vertebrae, typically less than 500 lb, result from studies of quasistatic loading, and large values, greater than 1000 lb, result from studies of dynamic loading. As the conditions of dynamic loading experiments are much more like manikin or aircrew member ejections than are quasistatic loading conditions, it is probably appropriate to use the larger ultimate strength data or the Hodgson-Thomas criteria, which accounts for duration of peak loading, in interpreting manikin test data.

It is certain that a manikin neck that is too simple will be incapable of predicting all of the types of failure that can occur in a human neck, and it is of particular importance that the range of validity of the manikin neck be established by comparison of results from tests with manikins and cadavers or, indirectly, by confirmation of proper manikin prediction of ejection-related injuries seen in operational conditions.

For various reasons adoption of conservative neck-injury criteria for trauma assessment in ejection studies is recommended.

AIRCREW EJECTION INJURY ANALYSIS

AND TRAUMA ASSESSMENT CRITERIA

1.0 BACKGROUND

The research reported here was conducted by the University of Michigan Research Institute (UMTRI) as a subcontractor to Conrad Technologies, Inc. This study, *Aircrew Ejection Injury Analysis and Trauma Assessment Criteria*, is one task of a larger research effort conducted by Conrad Technologies, Inc., for the Naval Air Warfare Center (NAWC) under Contract No. N62269-91-C-0225.

The overall goals of the Conrad Technologies research for NAWC relate to several aspects of increasing the safety of aircrew of Navy aircraft, both in normal operation of the aircraft and in emergency situations when it may be necessary to abandon the aircraft.

The specific goals of the research reported here are to document the types of injury that can occur during emergency egress from an aircraft by conducting a comprehensive literature review, and to perform an injury priority analysis and document criteria for trauma assessment. The results of this research are pertinent to the application of an ejection test manikin in Navy studies of automated escape systems.

2.0 APPROACH AND METHODS

Many different kinds of injury occur in association with aircrew member ejection from fighter and attack aircraft. The mechanisms of injury are diverse if "ejection" is considered to include all stages of emergency escape, i.e., from (or before) the activation of catapult ejection to recovery after landing. Injuries can result from all of the following elements of the escape and certainly from other contributing factors as well.

- aircraft maneuvering
- pre-escape positioning of crewmember
- ejection boost
- helmet impact with the canopy
- exposure to a rocket exhaust
- helmet windscoop and other windblast effects
- drogue opening shock
- parachute opening shock
- landing impacts
- rescue impacts

The scope of the current study is limited to the phases of the escape sequence that precede complete egress from the aircraft, i.e.,

- aircraft maneuvering
- pre-escape positioning of crewmember
- ejection boost
- helmet impact with the canopy
- exposure to a rocket exhaust

The three subtasks of this study are described in sections below. They are

Subtask 1 -- Literature Review
Subtask 2 -- Injury Priority Analysis
Subtask 3 -- Trauma Assessment Criteria

Subtask 1, the literature review, is the basis for both Subtask 2 and Subtask 3. In the Injury Priority Analysis subtask the most important observational ejection-related injuries are identified. This establishes the types of injury that are most important to study and, therefore, the types of dynamic response data that should be determined in manikin ejection tests. The Trauma Assessment Criteria subtask establishes how the dynamic response measures determined from testing should be interpreted in terms of injury potential.

3.0 LITERATURE REVIEW (Subtask 1)

A literature review was conducted for the purpose of documenting published information pertinent to the injury potential of events that can occur during the ejection phase of emergency escape from fighter and attack aircraft. The goals of the study are served by examining a large, if not complete, collection of the pertinent literature. Budgetary constraints, while making it impossible for an exhaustive compilation of the pertinent literature to be made, did not prevent the study from being major in scope. Papers, articles, and reports of the past ten years are considered of particular value, but all earlier, pertinent literature that was readily available at UMTRI was also reviewed.

Four types of references were obtained for review. The first type is comprised of references that provide statistical information relevant to incidence of ejection-related injuries of different sorts. Many of those papers, articles, and reports also have information regarding mechanisms of injury. These provide the basis for Table 8 of this report and for the findings for Subtask 2 (Injury Priority Analysis). The second type of reference focuses on the relationship between measures of dynamic response and the potential for injury--i.e., on injury criteria. These references also often contain (bio)mechanical property data for the elements of the human body related to particular kinds injury. This reference type provides the basis for Table 9 of this report and for the findings for Subtask 3 (Trauma Assessment Criteria). The third type of reference is comprised of documents that are pertinent to both Subtask 2 and Subtask 3. Information from those references is included in both Table 8 and Table 9. Finally, a fourth type of reference is comprised of documents that do not directly address the objectives of either Subtask 2 or Subtask 3, but are nonetheless pertinent to the overall goal of using ejection test manikins effectively to reduce the incidence and cost of ejection-related injuries. For the most part, the included references of this type relate to manikin design or to interpretation of test results so, in general, they are included in Table 9 (Subtask 3) but not in Table 8.

Although references of interest were identified, obtained, and reviewed throughout the course of the study, there were two primary stages of the procedure. The first stage sought to identify the injury types of greatest consequence; i.e., the literature review focused initially most closely on Subtask 2. The second primary stage of the literature review procedure, for Subtask 3, then focused on injury criteria for the types of injury identified in Subtask 2.

3.1 Impact Biomechanics. Much of the existing literature related to human tolerance to impact injury is relatively recent. The issue of automotive safety became a strong impetus for research in the field of impact biomechanics in about 1960, although some automotive company research, and perhaps a larger amount of military research in this field, was conducted as early as the 1930's. Automotive safety research that is related to design of restraint systems, occupant compartments, and crashworthy vehicle structures is more intensive now than ever. Similar military research is conducted for the purpose of protecting personnel not only in crash environments--the only situation of relevance in the case of automotive safety research --but also in high-G operational environments. The library of the University of Michigan Transportation Research Institute contains the largest collection in the world of references related to highway safety research--over 80,000 references in total and more than 200 journals and periodicals. Since information from nonclassified military research in the field of impact biomechanics is very pertinent to automobile occupant protection research, the UMTRI library contains a large number of references from Navy, Army, and Air Force research agencies.

3.2 Keyword Stems. The list of keyword stems below was used in computer searches of the UMTRI library data base. Searching the title field for these stems identified over one thousand references, most of which were eliminated as nonpertinent by reading the title. The references not eliminated on the basis of their titles were obtained and examined cursorily to determine the likelihood of their relevance to the study. Only a small percentage of these references was eliminated since in most cases the title was sufficiently descriptive of general content that nonpertinent references had already been eliminated. The references thus identified were supplemented by pertinent papers, articles, and reports found by examining all available AGARD reports and issues of *Aviation, Space, and Environmental Medicine*, which are available, but not completely indexed, in the UMTRI library data base. Most such references survived close review to be included in the final list of references. During compilation and analysis additional relevant documents were identified from the references of the documents under review. Some of these, too, were obtained and reviewed. All references obtained and reviewed by the process described here are in the List of References at the end of this report[1].

Table 1. KEYWORD STEMS FOR TITLES SEARCH

aircrew	eject	neck
axial	emergency	pilot
burn	escape	position
canopy	Gz	rocket
capsule	helmet	seat
catapult	impact	skull
cervical	injury	spinal
compression	intervertebral	spine
crew	jettison	thoraco
criteria	joint	tolerance
disc	ligament	trauma
disk	lumbar	vertebra
egress	manikin	

[1]The last item of each entry in the List of References is the UMTRI reference number, which has the form "UMTRI-nnnnn." Several references that were important to include in the list but are not available have "(unavailable)" in place of the UMTRI reference number. (Most of those are references cited in references in Tables 8 and 9.) Other references in the list have no UMTRI reference number but they are available and were reviewed; these are identified by "(no UMTRI number)." The List of References has 143 entries.

4.0 INJURY PRIORITY ANALYSIS (Subtask 2)

The overall goals of the Injury Priority Analysis subtask are to identify the most important observational, ejection-related injury types and to determine the effects of miscellaneous factors on the likelihood and severity of injuries.

Table 8 describes all papers, articles, and reports reviewed that are pertinent to Subtask 2. A synopsis of each reference is given in a "Summary/Comments" section, and the table also includes notations regarding escape sequence phases of pertinence, parameters addressed, and injury types addressed. Reference entries in the table are ordered inversely by date of publication since, in general, the references of greatest pertinence to the current state of emergency escape system effectiveness may be assumed to be the more recent ones.

A summary of the most important findings of Subtask 2 is given below in Sections 4.1 and 4.2.

4.1 <u>Injury Types and Injury Rates</u>. It is not always possible to associate injuries in specific ejections to exposures received during ejection, as opposed to post-ejection or even pre-ejection events. Further, Guill and Herd (1989a,b,c) suggest that many ejection-caused injuries, if not of a serious nature, go unreported by the aircrew member and undiagnosed by the attending flight physician. Additionally, they state that there is strong anecdotal evidence that, when coupled with ejection report data, suggest that a significant proportion of those ejectees sustaining an "ejection-associated" dynamic response-type neck injury might well have sustained their injury prior to the ejection, during the aircraft maneuvers and gyrations preceding the escape. Despite the certain presence of some amount of error in the data--possibly large--and also nonspecificities, which can make interpretations uncertain, it is clear from even a cursory review of the literature on ejection-related injuries that spinal column fractures are the dominant and most severe types of injuries that result during ejection-seat acceleration.

There is virtually no disagreement among the authors of the reviewed references regarding the most important injury types seen in ejection data bases, although it is noted here that the third finding below is based on fewer relevant references. Thus, the *central findings* of the current study as regards injury types and rates are these:

1) Fractures of lower thoracic and upper lumbar vertebrae, from T11 to L2, are the dominant major injuries that occur prior to complete egress from the aircraft during aircrew ejection. Such fractures occur in typically 20 percent of all ejections (7 to 47 percent, depending on the data base examined). Thoraco-lumbar fractures are most common at T12 and L1.

2) Fractures of the cervical vertebrae are five to seven times less common than fractures in the thoraco-lumbar spinal column, but they are nonetheless important since they are sometimes fatal and are much more often associated with permanent, major disability. Cervical fractures are most common at C2, C5, and C6.

3) Fatal head and neck injuries of nonspecific type may occur with significantly higher rates for through-the-canopy systems without fragmentation devices than for jettisoned-canopy systems.

While not expressing opinions contrary to these conclusions of the current study in any of their three 1989 papers, Guill and Herd add cautionary notes beyond those expressed above. They say (1989c) that determining the cause(s) for an ejectee's injuries is one of the more important and yet most difficult tasks associated with an ejection investigation. The authors argue that careful, detailed investigation (and also general statistical investigation) of ejection-associated injuries and circumstances often reveals that the assigned causal factors either cannot be applicable or are of extremely doubtful applicability for the specific situations. They argue, too, that aiding and abetting the selection of incorrect causal factors is the "strength-in-numbers" type of legitimacy that many factors have acquired through frequent usage over the years.

4.1.1 Fatal head and neck injuries. Head and neck fatal injuries related to the ejection procedure, the third item above, will now be discussed. This "injury type" is more specific with respect to cause than to type since approximately three-fourths of fatality injuries are typed as "multiple trauma" for each of the two types of escape systems--through-the-canopy (without canopy fragmentation devices) and jettisoned-canopy systems (Yacavone et al. 1992). Such fatal injuries are addressed by only four of the references reviewed--and in some cases only indirectly. Only one reference (Yacavone et al.) does more than relate fatality rates to the type of ejection system and make general comments regarding nonspecific kinds of head and neck injuries. The limited information found in the literature does, however, seem to show that fatal injuries to the head and/or neck can often be attributed to head impact with the canopy in through-the-canopy ejections. Yacavone finds that, compared with jettisoned-canopy ejections, through-the-canopy ejections result in fatality more than twice as often--10.7 percent to 4.7 percent. These rates include, however, the multiple-trauma fatalities that make up 70.3 and 77.7 percent of the fatalities, respectively. For the approximately 20 percent of fatalities that were attributed to skull-cervical fracture injuries (nearly all cervical) for both escape systems (22.2 and 19.4 percent), Yacavone indicates that there are significantly different causes. They state that there is a strong statistical association between fatal injury frequency and through-the-canopy ejections while for jettisoned-canopy ejections a greater proportion of fatalities result from striking part of the aircraft post egress. (Yacavone suggests use of canopy fragmentation explosive cords as a means of reducing forces on the aircrew member exiting through the canopy.) Guill and Herd (1989c) indicate higher rates of vertebral compression fractures in through-the-canopy ejectees. Data in Volume II of the Naval Safety Center reference (1981) show vertebral injury rates for through-the-canopy ejectees that are nine times as great as for canopy-jettisoned ejectees (September 1958 through December 1961). It is probably reasonable to infer from the Guill and Herd data and the Naval Safety Center data that fatal injury rates, too, are positively correlated with use of the through-the-canopy escape system. Contrarily, however, it must be noted that data in Volume IV of the Naval Safety Center reference, for all U.S. Navy ejections from January 1969 to December 1979 indicate cervical fracture rates of about 2 per-cent for both escape systems, i.e., no significant difference. Voge and Borowsky (1983) state that fractures and dislocations are the most common head and neck injury diagnoses in fatal ejections, occurring in 49 percent of the cases, but they do not associate those injuries with any specific cause.

If careful examination of all available data confirms a relationship between through-the-canopy ejections and increased likelihood of fatal head-neck injuries, the rates will still be small. Overall fatal-injury rates for fighter and attack aircraft ejections are less than 10 percent for modern escape systems and 10.7 percent, according to Yacavone for through-the-canopy systems (without fragmentation devices). (See Section 4.1.4 regarding fatality rates.) If 77.7 percent of those fatalities result from multiple trauma, as indicated by Yacavone, then no more than 22.3 percent--and probably much less--can be attributed directly to canopy versus head forces. This suggests that less than 2.4 of the 10.7 percent fatality rate for through-the-canopy systems could be attributed to canopy versus head forces. This is even larger than the 2 percent cervical spine fracture rates found in U.S. Navy ejection data for the period January 1969 to December 1979 (see Section 4.1.3), so the true rate is surely less than 2 percent and is, in all likelihood, a fraction of one percent. Still, a fatality rate of possibly one percent that is attributable to system design is unacceptably large if relatively simple implementation of canopy fragmentation explosive cords can significantly reduce the degree of hazard. (Also, see Chiou et al. (1993) regarding canopy fragmentation with MDCs, i.e., miniature detonating cords.)

4.1.2 Thoraco-lumbar fractures. Regarding rates of occurrence of thoraco-lumbar fractures, Visuri and Aho (1992) indicate a 19 percent occurrence among ejection survivors. Sandstedt (1989) indicates 18, 27, 21, or 20 percent for various combinations of sitting posture and flight condition. McCarthy (1988) determines a rate of 21 percent for survivors in takeoff-and-landing ejections and a similar result for ejections above 500 feet altitude; major injury rates, including thoraco-lumbar fractures, are two and a half times as great as this for non-takeoff-and-landing ejections below 500 feet. Data for U.S. Navy ejections that occurred from January 1969 to December 1979 were analyzed and presented at the 1981 symposium sponsored by the Naval Safety Center. Volume IV of that reference indicates a thoraco-lumbar fracture rate of 28 percent in through-the-canopy ejections but only 7 percent in jettisoned-canopy ejections--10 percent overall for 1120 ejections. In a somewhat earlier study by Auffret and Delahaye (1975) spinal fractures were found to occur in 10 to 47 percent of surviving ejectees depending on the data base examined; 37 percent of all fractures occurred at T12 or L1. Rotondo (1975) finds a 36 percent occurrence rate among survivors of Italian pilot ejections. The distribution of fractures was nearly uniform over the entire range of occurrence, from T7 to L4, except for T12 and L1, where the rate of occurrence was nearly four times as great. Nuttall (1971) identifies T11 to L2 as the part of the spine where fracture is most likely. Regarding rate of occurrence of spinal fractures, however, he cites a 1957 study and a 1965 study that determined that only 2.2 percent and 3.8 percent, respectively, of ejections cause spinal fractures. Symeonides (1971) indicates an 18 percent spinal fracture rate among surviving ejectees; most fractures were in the T11 to L2 region. Henzel (1967) indicates T10 to L1 as the most common location of fractures. Jones (1964) found that T12 was the most common injury site and that L1 was the most common lumbar injury site.

Regardless of the region of the spinal column considered--thoracic, lumbar, or cervical--the occurring fracture injuries are predominantly anterior-lip crush fractures that result from hyperflexion (e.g., Naval Safety Center, Vol. II, 1981; Kazarian et al. 1979; Kazarian, 1978; Auffret and Delahaye, 1975; Chen, 1973 [simulation]; Ewing et al. 1973 [experimental]; Nuttall, 1971; Shannon, 1971).

4.1.3 Cervical fractures. As stated above, fractures of the cervical vertebrae are found to be five to seven times less common than fractures in the thoraco-lumbar spinal column. Guill and Herd (1989a) indicate a very low rate of cervical fractures among survivors of U.S. Navy ejections from 1949 to 1968--just 12 in 1764 ejections. For the period 1969 to 1988 they indicate 28 in 1677 ejections--less than 2 percent. Their data in another paper (1989c), for nonfatal injuries attributed to ejection, show an occurrence rate that is seven times greater for thoraco-lumbar fractures than for cervical fractures. Voge and Borowsky (1983) determined that in nonfatal ejection incidents in which vertebral fracture(s) occurred, 81 percent of ejectees had thoraco-lumbar fractures and 13 percent had cervical fractures--a ratio of six to one. Volume IV of the Naval Safety Center reference mentioned above indicates a cervical fracture rate of 2 percent in both through-the-canopy and jettisoned-canopy ejections. The corresponding rates for thoraco-lumbar fractures are fourteen and four times as great, with an overall ratio of five to one relative to cervical fracture rates. Zenobi (1978) states that

U.S. Navy data from ejections during 1967 to 1974 show that neck injuries ranging from minor to critical occurred at a rate of approximately 8 percent. A study by Guill and Herd (1989a) indicates that, among survivors, cervical sprain or strain is seven to eleven times as common as cervical fractures, so the implied cervical fracture rate in the Zenobi study is about one percent.

Guill and Herd (1989a,b) express the opinion that there is no single, primary causal factor for serious neck injuries, but that, rather, the underlying causal factors are many and varied. They find that neck injuries associated with ejections do not conform to the patterns expected for any single proposed causal factor and mechanism that have been advanced to date. They believe additionally, however, that there is evidence that many reported neck injuries are the consequence of system malfunction, e.g., the seat striking the ejectee during parachute opening following man-seat separation and the entanglement of the ejectee with the seat prior to parachute opening.

4.1.4 Fatal injuries: general. Approximately three-quarters of fatal injuries in ejections are the result of multiple trauma according to Yacavone et al. (1992). Such injuries result largely from forces other than ones experienced by the ejectee prior to complete egress from the aircraft so, by the defined scope of the current study, general fatality statistics are not relevant here. Nonetheless, a summary of statistics from the reviewed literature is presented. Fatalities here are from all causes, ejection related or not. (In general the reviewed references do not attempt to describe the various associated factors statistically.) Visuri and Aho (1992) find a fatality rate of 5.9 percent for a small data base (17 ejections). Yacavone et al., as discussed in Section 4.1.1, find a rate of 10.7 percent for through-the-canopy ejections and 4.7 percent for jettisoned-canopy ejections. Guill and Herd (1989c) state a rate of 15 percent for U.S. Navy ejections from 1949 to 1982. Sandstedt (1989) data show a rate of 9.8 percent for 92 ejections. McCarthy (1988), in a study of takeoff-and-landing ejections, finds an overall fatality rate of 13.7 percent and a rate of 11.5 percent for ejections above 500 feet. Non-takeoff-and-landing ejections below 500 feet have a 53.7 percent associated fatality rate. The fatality rate in 1967-1980 ejections studied by Hearon et al. (1981) was 20 percent. In a study of U.S. Air Force ejections in 1968-1970, Shannon (1971) determined a fatality rate of 11 percent.

4.1.5 Windblast and parachute opening shock injuries. While not of direct relevance to the goals of this study as stated, it is nonetheless important to comment on the prevalence of the primary injury causations not dealt with--windblast and parachute opening shock--relative to rates for the major injury types identified in the study. Windblast injuries are of various types, including (primarily) limb flail. Parachute opening shock and ground impact can produce significant $+G_z$ forces, although, because of different constraints, they produce different injury patterns. Brinkley and Shaffer (1971) state that ejection boost acceleration is the primary cause of major injuries related to ejections--84 percent in a study of F-4 ejections--and that the second largest cause is post-ejection limb flailing, which accounts for 12 percent of the total number of major injuries. In their study they found that only five major injuries resulted from parachute opening shock in 384 ejections (1.3 percent). Shannon (1971) determined in a U.S. Air Force study that in 62 percent of major-injury cases, ejection forces were judged responsible for the primary injuries. Windblast and parachute opening shock were identified in 28 percent and 10 percent, respectively, of the cases.

4.2 Effects of Influencing Parameters. Many factors besides ejection boost forces affect the performance of an ejection escape system. The influence of various factors on injury rates is discussed in many of the references that are mentioned in the previous section and summarized in Table 8. The primary parameters currently thought to be of possible importance are: type of aircraft maneuver, crewmember pre-escape positioning, ejection boost forces, helmet impact with the canopy, aircraft speed, severity of maneuvers, mission requirements, and crewmember physiology and anthropometry.

Information from the references that is relevant to these factors is summarized in the following subsections. Although much of the most important information is presented below, this tabulation does not cover all relevant material in the references. Detail of interest can be found by referencing Table 8 and consulting the documents. (See Desjardins et al. (1982) in addition to references mentioned below.)

4.2.1 Type of aircraft maneuver.

Hämäläinen and Vanharanta (1992)

-- High performance maneuvers such as in combat can result in neck muscle strains as great as 5.9 times strains at 1.0 G_Z and 37.9 percent of the maximal voluntary contraction (MVC). Pilots in the study experienced severe neck pain at $+G_Z$s of much less than ejection boost accelerations.

Guill and Herd (1989a,b)

-- A significant proportion of the serious ejection-associated neck injuries are likely to have been induced by the inflight maneuvering/gyration forces imposed upon the aircrew prior to ejection or during ejection. Nonetheless, this is not a *primary* factor in explaining ejection-related *neck* injuries.

McCarthy (1988)

-- Fatality and major-injury rates for ejections during takeoffs and landings are very little different from rates for ejections from above 500 feet.

Higgins et al. (1965)
- -- High performance maneuvering is detrimental to the ejection success rate to the extent that it might cause the aircrew member to be out of position (not erectly seated) during ejection.

4.2.2 <u>Crewmember pre-escape positioning</u>.

Freivalds and McCauley (1990)
- -- Ejection simulations show that head and neck angles during catapult boost need to be aligned and vertical to reduce neck flexion torques. Added helmet mass has little effect on the likely severity of injury due to the $+G_Z$ acceleration if head and neck position is proper.

Guill and Herd (1989a)
- -- Poor body position is not a primary factor in explaining ejection-related neck injuries.

Naval Safety Center, Vol. II (1981)
- -- Most ejection-associated, vertebral-compression fractures are the result of poor vertebral alignment. Causes include personal equipment influences, nonstable ejection platform, inadequate thigh support, poor torso restraint, forward torso rotation induced by rear-angled catapult boost acceleration vector, poor seatback support, and upper torso movement.
- -- Equipping ESCAPAC seats with powered inertia reels to force the ejectee into a torso-back, erectly seated position prior to ejection boost reduced the rate of lower thoracic and upper lumbar fractures by a factor of two and reduced the rate of neck sprain/strain by a factor of six. It increased the rate of cervical and midthoracic fractures.
- -- The primary negative influence of head-canopy contact in through-the-canopy ejections may be the inducement of vertebral misalignment.

Fleming (1979)
- -- It is better to use an upper ejection handle than a low ejection handle because it allows the ejectee to maintain a more erect seated position.

Kazarian et al. (1979; 1977)
- -- Midthoracic fracture rates are much greater when a powered inertial reel is used. While it reduces lower thoracic and upper lumbar fractures, it causes pre-ejection midthoracic hyperflexion by powerfully forcing the torso back against the seat. These injuries are a function of seat geometry and harness configuration.

Kazarian (1978)

-- Upper and midthoracic hyperflexion and hyperextension injuries are induced by powered inertial reels. Individual torso height and restraint system geometry are factors.

Auffret and Delahaye (1975)

-- The most important factors affecting likelihood of injury are the posture and position of the pilot at the moment of ejection. The pilot should be seated erectly and should be restrained by a harness that does not allow excessive freedom of movement of the torso (especially in flexion). The harness should be tight enough to hold the pilot in position even in high-G maneuvers since abnormal flight configurations may well exist at the instant of ejection. The seat pan angle should be such that the angle between the torso and the thigh is 135 degrees for proper alignment of the thoracic vertebrae.

Nuttall (1971)

-- An erect posture with the head and buttocks pressed firmly back into the seat is an important factor in preventing spinal fractures. Fracture rates can be as much as 13 times greater for improperly positioned ejectees.

Shannon (1971)

-- The spinal fracture rate for optimally seated ejectees (head and buttocks back into the seat) was 4 percent; the rate for improperly seated ejectees was 31 percent, i.e., eight times as large.

Symeonides (1971)

-- Tightening the shoulder-buttock belts excessively can force the shoulders down and cause a preflexed state of the spine, increasing its vulnerability to $+G_Z$ forces.

Higgins et al. (1965)

-- Proper body position and execution of ejection procedures reduces spinal fracture rates.

4.2.3 Ejection boost forces. There is much in the literature regarding the effects of various parameters of the catapult and ejectee acceleration profiles. These parameters include peak $+G_Z$ acceleration, the rate of onset of the acceleration profile, and velocity at end of stroke. These parameters will not be addressed here except to say that peak acceleration magnitudes of 20-25 G, rates of onset of 200-500 G/s, end-of-stroke velocities of less than 20-60 ft/s (depending on system), and stroke durations of 230 ms or more are generally believed to be noninjury producing, provided that the ejectee's spinal column is properly aligned. Information related to these factors may be found in Table 9 and in Section 5.0.

Brinkley and Shaffer (1971)

-- It is important for the catapult acceleration vector to be aligned with the crew member's vertebral column to reduce the occurrence of spinal fractures.

Nuttall (1971)

-- The thrust vector of the seat should be parallel to the spinal column or forward from it to prevent anterior-lip compression fractures in the cervical and upper-thoracic spine.

Shannon (1971)

-- U.S. Air Force data from 1968 to 1970 show that the major-injury rate for straight ballistic catapult systems was 12 percent for all nonfatal ejectees; the rate for rocket-assisted systems was 8 percent.

Higgins et al. (1965)

-- The ejection axis should be parallel to the spinal axis.

4.2.4 Helmet impact with the canopy.

Chiou et al. (1993)

-- Reducing the probability of spinal injury is still the main concern in escape ejections. Canopy fragmentation through use of MDCs (miniature detonating cords) is an effective way to accomplish this for through-the-canopy ejections.

Yacavone et al. (1992)

-- U.S. Navy ejection data for the period 1977 to 1990 show that through-the-canopy ejections have higher associated injury rates than canopy-jettisoned ejections. Comparative rates are: fatalities, 10.7% vs. 4.7%; one work day lost, 29.2% vs. 17.4%

Guill and Herd (1989a)

-- Canopy mode is not a primary factor in explaining ejection-related neck injuries.

Naval Safety Center, Vol. II (1981)

-- Induced vertebral misalignment as well as head-canopy forces are factors in increased vertebral fracture rates in through-the-canopy ejections.

Naval Safety Center, Vol. IV (1981)

-- It may be determined from presented U.S. Navy ejection data for the period 1969 to 1979 that there are in excess of 2.5 injuries (minor and major) per ejectee in through-the-canopy ejections. For jettisoned-canopy ejections the average number of injuries is about 1.2--i.e., about half the rate for through-the-canopy ejections. There was no significant difference seen in fracture rates for the cervical spine.

Norman et al. (1979)
- In through-the-canopy ejections a double hit of the head against the canopy greatly reduced the protection provided by all types of helmets tested.

4.2.5 Aircraft speed.

Chiou et al. (1993)
- For all air speeds for which canopy fragmentation was tested, from 0 to 600 knots, it was found that the likelihood of injury to aircrew members from sharp edges of fragments or from impact by pellets of the lead skin of MDCs is not significant.

Guill and Herd (1989a)
- Aircraft speed (ejection air speed) is not a primary factor in explaining ejection-related neck injuries.

McCarthy (1988)
- Fatality and major-injury rates for ejections during takeoffs and landings (i.e., relatively low aircraft speeds) are very little different from rates for all ejections from above 500 feet (higher aircraft speeds).

Higgins et al. (1965)
- Fatality is more likely for ejections at aircraft speeds above 500 kn than at speeds below 500 kn.

4.2.6 Severity of maneuvers.

Hämäläinen and Vanharanta (1992)
- Under +7.0 G_z in bank maneuvers, neck muscle strains nearly six times the strains at 1.0 G_z were measured. One hundred percent of muscular tolerance was reached at +4.0 G_z in some high-severity maneuvers. Pilots in the study experienced severe neck pain at $+G_z$s of much less than ejection boost accelerations.

Guill and Herd (1989a)

- Severity of aircraft maneuvers is not a primary factor in explaining ejection-related neck injuries.

4.2.7 Mission requirements. Mission requirements were not addressed in specific terms in any of the references reviewed. Pertinent aspects of mission requirements, however, include: (1) necessity of high-severity maneuvers in combat (see Section 4.2.6); and (2) mission duration. Mission duration is a factor in neck fatigue and, therefore, is also a factor in cervical injury probability if an ejection is required. (See Hämäläinen, 1993, and Phillips and Petrofsky, 1983, in Table 9.)

4.2.8 Crewmember physiology and anthropometry.

Visuri and Aho (1992)
- No statistically significant relationships were found between ejection injury rates and height-weight index or age of the ejectees.

Guill and Herd (1989a)
- Ejectee anthropometry and preexisting neck injuries are not primary factors in explaining ejection-related neck injuries.

Hearon et al. (1981)
- Spinal injury rate can be a function of ejectee seated height if the restraint system does not allow adjustment of the shoulder harness angle.

Kazarian (1978)
- Individual torso height may be a factor in midthoracic injury rates associated with use of powered inertial reels.

Rotondo (1975)
- Individual preexisting spinal conditions are probably factors in the likelihood of spinal injury in an ejection. These include lumbago, discal prolapse, arthrosis, ischialgia, kyphotic and scoliotic deviations, spondylolysis, and spondylolisthesis.

Shannon (1971)
- No significant differences were found in injury rates for crewmembers of different weights when other factors, such as prepositioning, were considered.

Henzel (1967)
- Many ejection-incurred spinal injuries may result from unsuspected, congenital spinal weakness.

Higgins et al. (1965)
- There is no evidence of a relationship between ejectee height/weight and likelihood of injury. Pilots over 24 years of age are more likely to incur vertebral injury than younger pilots.

5.0 TRAUMA ASSESSMENT CRITERIA (Subtask 3)

The overall goal of the Trauma Assessment Criteria subtask is to develop a basis for relating dynamic response parameters that can be measured with manikins under experimental conditions to injuries that may be sustained by an aircrew member in a real-world ejection. Specifically, data are sought that are pertinent to assessment of potential for injuries of the types identified in Subtask 2, viz., fracture injuries of the thoraco-lumbar and cervical regions of the spinal column.

Table 9 describes all papers, articles, and reports reviewed that are relevant to Subtask 3. A synopsis of each reference is given in a "Summary/Comments" section, and the table also includes notations regarding injury criteria and biomechanical properties addressed. Reference entries in the table are ordered inversely by date of publication.

A summary of the most important findings of Subtask 3 is given below in Sections 5.1 and 5.2. It is beyond the scope of the current study to include more than a few graphical representations of dynamic response data, injury criteria curves, etc., that are in many of the references and relate to the objectives of Subtask 3. A written description of such data, with numerical values, is given here. The relevant references should be consulted if greater detail is needed.

5.1 <u>Injury Criteria</u>. For the purpose of this study injury criteria are considered to be relationships between measures of mechanical loading (or conditions of loading) and the levels of (or probability of) resulting injury. There are three fundamentally different types of biomechanical injury criteria in general use. Since, ultimately, all injuries occur at a cellular level, the first commonly used type relates stresses and strains in *tissues* to tissue injury. Instead of stresses and strains, the second type considers the gross characteristics of response--e.g., forces, moments, accelerations, etc.--of *elements of the human body* in characterization of injury probability or severity. The third type considers gross characteristics of dynamic loading of *the human body as a whole*.

In general, it is not possible to predict the probability or severity of major injuries to the living human being by measuring stresses and strains in tissues, whether in cadavers or in volunteer subjects. The experimental difficulties involved are obvious; they relate to measuring tissue stresses and strains, to assessing degree of injury at a cellular or tissue level, and to relating tissue-level injury to clinically observed injury--i.e., body element-level injury observed or diagnosed for living human beings. Additionally, however, it is the consensus that tissue-level injury criteria--even if obtainable--are not, in general, of practical use in understanding injury mechanisms. Melvin (1979) states, for example, that because of the complex structural interactions that can occur between the components of the neck, it is necessary to define injury criteria in terms of forces and moments acting on the neck, rather than the stresses and strains in the tissues that are actually damaged (e.g., the spinal cord, laryngeal cartilages, etc.). Apart from such considerations, as a practical matter it is not possible currently to construct test manikins capable of accurately simulating and measuring all of the pertinent tissue-level stresses and strains that might occur in a living human being in high-G impacts.

Thus, there are two types of injury criteria considered in this study. They relate injury to gross measures of response or loading--forces, moments, accelerations, etc.--of, respectively, (1) body elements and (2) the human body as a whole. For both thoraco-lumbar and cervical fractures these two types are discussed below in subsections 5.1.1.1, "Moment, force, and dynamic response criteria," and 5.1.1.2, "Ejection seat dynamics criteria."

5.1.1 Thoraco-lumbar spine fractures. Fracture of thoraco-lumbar spinal vertebrae, particularly from T11 to L2, is the most common major injury that occurs in ejections before complete egress from the aircraft. Useful information about the effectiveness of an ejection system design can be obtained even from test manikins that are capable of measuring only whole-body responses--e.g., thorax center-of-gravity $+G_Z$ as a function of time. The corresponding probability of fracture injury can be estimated by making use of observational injury data together with operational ejection system parameters. However, for a test manikin to be discriminating enough to predict specific injuries and injury mechanisms, it clearly must be capable of measuring appropriate body-element responses at primary injury sites--e.g., compressive anterior- and posterior-lip loads at "T12/L1" of the manikin.

5.1.1.1 Moment, force, and dynamic response criteria.

Moment and force criteria--No moment-related injury criteria for the thoraco-lumbar spine were found in references reviewed in the current study. Numerous researchers (e.g., Stech, 1963; Payne, 1971; Coltman et al. 1986), however, have found that the ratio of spinal compressive load to vertebral ultimate failure load is a good indicator of the potential for spinal injury. That is, with a reasonable amount of consistency, the probability of compressive fracture for any vertebra can be predicted from the compressive load. This assumes two things: first, that a statistically sufficient amount of compressive strength data are available from tests with cadaveric preparations for the particular vertebral level (C1 to L5) or that scaling between levels can be demonstrated to be valid and, second, that "probability" for a fracture is adequately defined. As they relate to the current study, there are minor problems with both of these. First, while a number of authors present experimental compressive strength data for vertebrae, most available data are for materials from cadavers of age 60 and above whereas most fighter and attack aircraft pilots are (males) in their 20's and 30's. However, while compressive strength does vary with biological age (Stech and Payne, 1963; Henzel, 1967; Payne, 1971), it is nearly independent of (adult) age for ages less than about 42 years (Payne, 1971). Data most useful for the current study will be from authors who use materials from young adult male cadavers (or properly adjust data from older cadavers). With regard to defining the probability for fracture, in most studies there are insufficient cadaveric test data to do more than either define conservative fracture strengths or median fracture strengths. Cadaveric test data normally do not permit meaningful definition of a fracture probability curve as a function of compression force. Rather, more simply, a conservative ultimate strength might be defined as the upper limit of compression force values for which *almost all* specimens do not fracture. A median strength might be defined as a value of compression force above which *approximately half* of specimens fail. (Alternatively, a mean, i.e., average, strength might be used.)

These caveats notwithstanding, consistent and useful data are found in the literature. Four reports and papers reviewed in the current study include ultimate compressive strength data by level for thoraco-lumbar vertebrae (mostly T1 to L5). Coltman et al. (1986) give data from tests of vertebrae from 12 cadavers. The ages ranged from 44 to 63 years (average, 56.25); eight of the 12 were male. Kazarian and von Gierke (1978) give data from tests for fast and slow loading rates (0.889 and 0.0000889 m/s), but they do not give information regarding age of the cadaver(s) from which vertebral specimens were taken or the number of cadavers used. Data presented by the authors from other researchers are bracketed by their data for fast and slow loading rates. Payne (1971) gives compressive strength data for levels C4 to L1 for one 30-year old male (Messerer, 1880) and for levels T8 to L5 for ten adult cadavers with an age range of 19 to 46 years and average 32.4 (Geertz, 1946 translation). He also gives a curve that shows cumulative probability of compressive failure as a function of load (adjustment of data from Bell et al. 1967). The data are normalized to L5 and age 42.5 years and are based on tests of 62 vertebral bodies. Henzel (1967) gives compressive strength data due to Ruff (1950), Stech (1963), and Perey (1957) for T1 to L5; all data are for young adult males.

Compressive strength data from these articles and reports are given in Table 2 below. Since all authors find that in good approximation the ultimate compressive strengths of thoraco-lumbar vertebrae increase linearly, by level, from T1 to L5, their results have been summarized in equation form in the table. (All authors presented their results in tabular and/or graphical form.) Here, L=1 for T1, L=2 for T2, ..., and L=17 for L5. Since the original data are variously in terms of pounds, Newtons, and kilograms force, some results have been converted to pounds for ease in comparison. (Values in the authors' original units may be found in the respective Table 9 entries.)

Table 2. THORACO-LUMBAR VERTEBRAL STRENGTH BY LEVEL				
Article / Report	**Vertebral Levels**	**Strength (S) by Level (L)**	**Cadavers n**	**Cadavers age**
Coltman et al. (1986)	T1-L5	AVERAGE	12	44-63
		S(lb) = 335 + (L-1)*(2015-335)/16		
		slope = 105 lb/level		
		(GREATEST)	1	52
		S(lb) = 1193 + (L-1)*(3881-1193)/16		
		slope = 168 lb/level		
		(LEAST BOUND)	≈5	54-63
		S(lb) = 200 + (L-1)*(1400-200)/16		
		slope = 75 lb/level		
Kazarian and von Gierke (1978)	T1-L5	AVERAGE, fast loading	?	?
		S(lb) = 719 + (L-1)*(3170-719)/16		
		slope = 153 lb/level		
	T1-L5	AVERAGE, slow loading	?	?
fast = 0.889 m/s		S(lb) = 562 + (L-1)*(1439-562)/16		
slow = 0.0000889 m/s		slope = 55 lb/level		
Payne (1971)	C4-L1	One male cadaver	1	30
		S(lb) = 606 + (L+3)*(2205-606)/16		
		slope = 100 lb/level		
	T8-L5	Ten adult cadavers	10	19-46
C4-L1, Messerer		S(lb) = 1357 + (L-8)*(2341-1357)/9		
T8-L5, Geertz+		slope = 109 lb/level		
Henzel (1967)	T1-T5	AVERAGE	?	young adult
		S(lb) = 360 + (L-1)*(840-360)/4		
		slope = 120 lb/level		
	T6-T10	AVERAGE		
		S(lb) = 1000 + (L-6)*(1632-1000)/4		
		slope = 158 lb/level		
	T11-L1	AVERAGE		
		S(lb) = 1700 + (L-11)*(1790-1700)/2		
		slope = 45 lb/level		
	L2-L5	AVERAGE		
		S(lb) = 1925 + (L-14)*(2366-1925)/3		
		slope = 147 lb/level		

NOTES:

Vertebral levels
L=1 for T1, L=2 for T2, ..., and L=17 for L5
L=-3 for C4, L=-2 for C5, L=-1 for C6, L=0 for C7

Constants for force units conversion
1 lb = 4.44822 N
1 kgf = 2.2046 lb

Units of original data

Coltman et al.	lb
Kazarian and von Gierke	Newtons
Payne	kgf
Henzel	lb

It may be seen in this strength-versus-level table that there is good general agreement between the slope values from data from Coltman et al., Kazarian and von Gierke, Henzel, and Payne (from Messerer and Geertz). Further, load values calculated from the equations for the various authors are similar. For example, for T8(L=8) the following loads are calculated: 1070 lb for Coltman et al., "AVERAGE"; 1547 lb for Coltman et al., "avg GREATEST+LEAST"; 1369 lb for Kazarian and von Gierke, "avg fast+slow"; 1705 lb for Payne, "Messerer"; 1357 lb for Payne, "Geertz"; 1200, 1316, 1565, and 1043 lb for Henzel, "AVERAGE". (The last two values for Henzel are extrapolations to T8.) Corresponding slope values are, respectively, 105, 122, 104, 100, 109, 120, 158, 45 (T11 to L1 only; Henzel), and 147 lb/level.

Only one reference reviewed in the current study contains information that describes the probability of vertebral body fracture as a function of compressive load--viz., Payne (1971). Payne examines data from Geertz (1946), Perey (1957), and Bell et al. (1967). The pertinent analysis and results will now be described.

Payne looks at the relationship between ultimate strength and vertebral level. For this purpose he finds only the Geertz data plus three data points from another source to be useful. Using data from 38 vertebral bodies between T8 and L5 from ten cadavers (age 19 to 46) he finds the relationship to be linear. Payne's plotted points and his regression line are shown in Figure 1. (This is Payne's Figure 19.) Payne does not note the values of the regression line parameters in his paper, but they may be calculated to be as follows (using the data in Payne's Table 2), where S is the ultimate compressive strength and L is vertebral level from 8 to 17:

$$S(\text{kgf}) = 615.465 + (L-8)*(1061.728-615.465)/9,$$

where L=8 for T8 and L=17 for L5

slope = 49.585 kgf/level
correlation coefficient = r = 0.8367
standard error of estimate of S on L = 87.444 kgf
standard deviation of S = 159.66 kgf

The above regression line equation is equivalent to the one in the above table (viz., Payne, T8-L5), where results are expressed in pounds:

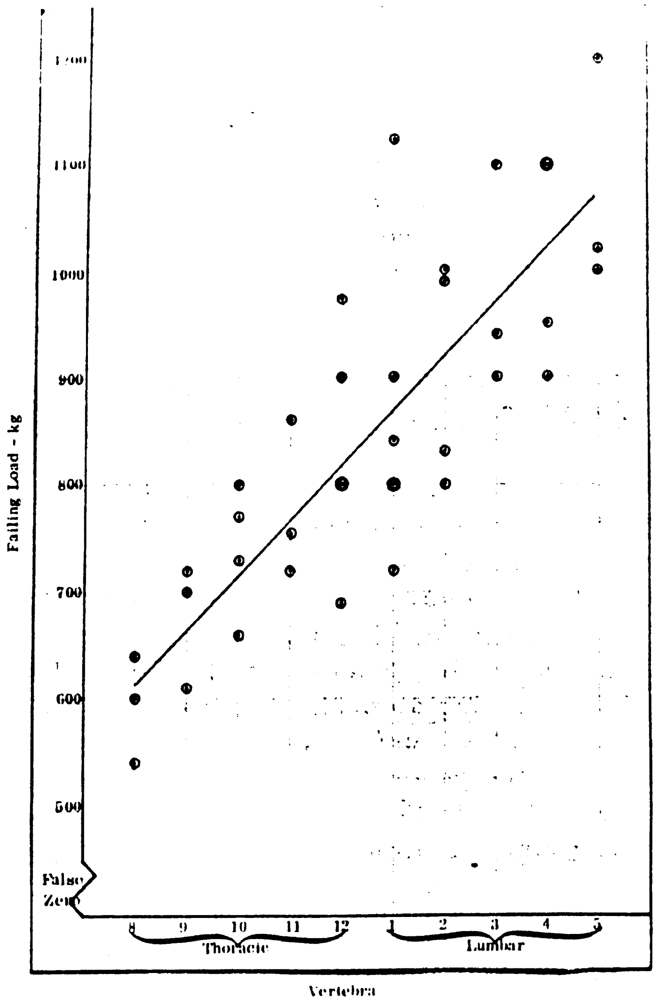
$$S(\text{lb}) = 1357 + (L-8)*(2341-1357)/9$$

slope = 109 lb/level

Payne next determines, from analysis of two sets of data, that compressive breaking load is independent of age up to about 42 years and that it decreases exponentially above that (Payne, Fig. 28). Indeed, Payne states explicitly that "as a practical matter, we may neglect the effect of age when considering the

Figure 1.

Vertebral failing load, T8 to L5 (from Payne, 1971)



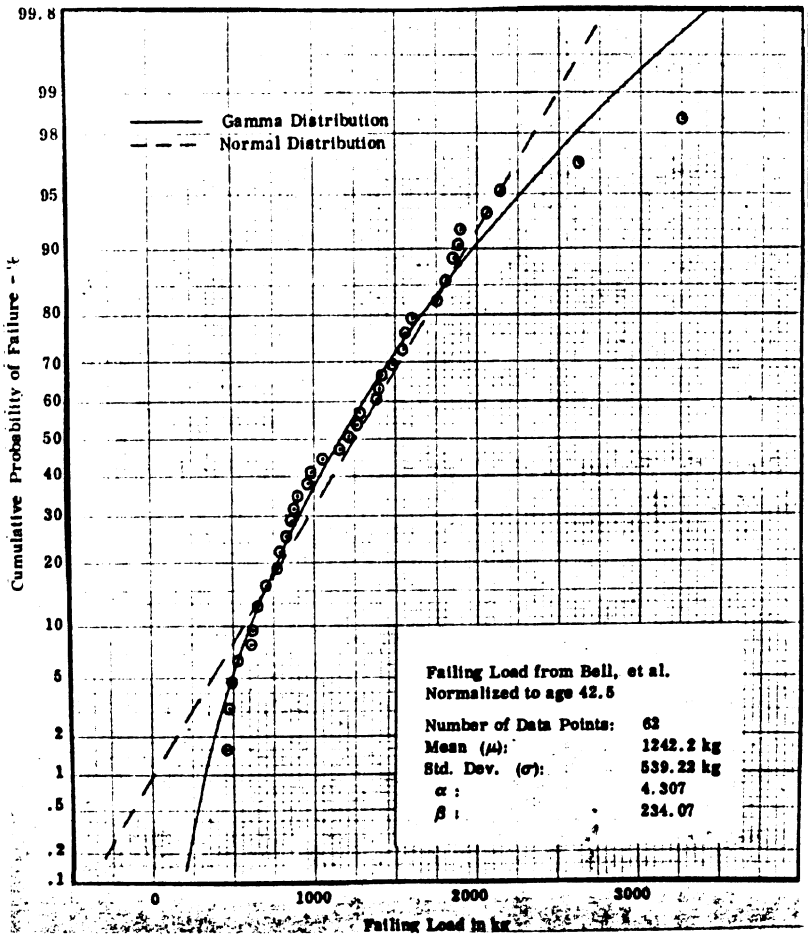


Figure 2.

Vertebral strength, normalized to age 42.5 and based on 2.77 in² L5-body area (from Payne, 1971) problem of aircrew injury in ejection seats." For ages above 42.5 years he determines parameter values for the best-fit exponential relationship, which he uses to normalize to age 42.5 the data from Bell et al., which are for cadavers of age 26 to 86. Since Bell's data, for 62 thoraco-lumbar vertebrae, are for compressive stresses instead of compressive loads, Payne uses vertebral body cross-sectional areas (2.77 in^2 for L5) to find equivalent loads and to normalize to L5. The cumulative probability-of-failure relationship he determines from his analysis of Bell's data is shown in Figure 2 (Payne's Figure 35). The plotted data are normalized to age 42.5 and to L5; i.e., they may be considered valid for ages less than 42.5 years (since he finds strength to be independent of age less than 42.5) and they are for L5 specifically although data for a range of ages and for vertebral levels other than L5 were used to establish the results. Payne finds that a gamma distribution fits the data well. He gives the following equation for the best-fit curve for the probability density function, which has units of probability per kgf:

$$p(S) = \frac{S^{\alpha} e^{-S/\beta}}{\beta^{\alpha+1}\,\Gamma(\alpha+1)}, \qquad \alpha > -1, \quad \beta > 0$$

where $\alpha = 4.307$ and $\beta = 234.07$ kgf, and S is the L5 load in units of kilograms force. The factor $\Gamma(\cdot)$ is the gamma function. This equation is most conveniently used in a slightly different form:

$$p(S) = \frac{(S/\beta)^{\alpha} e^{-S/\beta}}{\beta\,\Gamma(\alpha+1)}, \qquad \alpha > -1, \quad \beta > 0 ,$$

where $$\Gamma(\alpha+1) = \int_0^{\infty} e^{-u} u^{\alpha}\, du = 26.1428 \quad \text{for } \alpha = 4.307.$$

The cumulative probability-of-failure curve of Figure 2 is then

$$P(S) = \int_0^{S} p(u)\, du ,$$

which can be shown to be

CUMULATIVE PROBABILITY OF FAILURE FOR L5 COMPRESSIVE FORCE S

$$P(S) = \frac{1}{\Gamma(\alpha+1)} \int_0^{S/\beta} e^{-t}\, t^{\alpha}\, dt\,, \quad \alpha > -1, \quad \beta > 0$$

where α, β, and $\Gamma(\alpha+1)$ have the values given above.

Payne does not describe the manner in which his results might be used to estimate the probability of failure of a thoraco-lumbar vertebra at a particular level L for a given maximum (quasistatic) compressive load F measured at that level. It would seem, however, that the proper procedure is as follows. Given a level L, where L=8 for T8, 9 for T9, ..., 12 for T12, 13 for L1, ..., and 17 for L5, calculate the estimated ultimate compressive strength, S, from

$$S_L(\text{kgf}) = 615.465 + (L-8)*\text{slope}$$

where

$$\text{slope} = 49.585 \text{ kgf/level}$$

and where loads are in units of kilograms force. This equation is for the regression line in Figure 1. Since Payne's cumulative probability-of-failure curve is normalized to L5, we also need the strength for L5. For L5 (L=17) we have

$$S_{17} = 1061.728 \text{ kgf}\ .$$

Next, for the measured value of load at level L, i.e., F, calculate the ratio R of load to the estimated, nominal breaking strength:

$$R = \frac{F}{S_L}$$

The equivalent load at L5 may then be determined as

$$F_{L5} = R\ S_{17}\ .$$

Finally, with S in the above equation for P(S) set to F_{L5}, the cumulative probability of failure for loads up to F_{L5} at level 17 (L5)--and, equivalently, F at level L--may be calculated. Alternatively, the probability may be read directly from Figure 2 for abscissa value F_{L5}.

To illustrate an inverse use of the above procedure we may note that Figure 2 shows that 25, 50, and 90 percent probabilities of failure of L5 occur at L5 loads of about 800, 1150, and 2000 kgf, respectively. For T10 (L=10) the nominal breaking load is found to be S_{10} = 714.6 kgf so that the ratio R is 714.6/1061.728, or 0.673. The 25, 50, and 90 percent probabilities of failure of T10 therefore occur at T10 loads of about 538, 774, and 1346 kgf, respectively.

One additional, and possibly important, caveat must be expressed regarding prediction of thoraco-lumbar vertebral fracture. All or almost all ultimate strength data in the literature for T1 to L5 were determined from experiments with loading rates that are small in comparison with loading rates during ejections. Yet there is indication that ultimate strengths for high loading rates may be significantly larger. As seen in Table 2, for example, Kazarian and von Gierke get a T1-strength value of 562 lb for quasistatic loading but 719 lb for a loading rate of about 1 m/s, i.e., a strength that is larger by 28 percent. A much larger amount of dynamic loading data exists in the literature for compressive strength of vertebrae in the cervical spine. Those data exist because of a strong focus in automotive safety research on neck injuries. Maximum loading rates studied are usually about 10 m/s. The related literature is discussed in Section 5.1.2.1. It is seen there that cervical vertebra strengths can be two to three times as large, and more, in dynamic loading as in quasistatic loading. (Thoraco-lumbar vertebra strengths have not received much attention in automotive safety research because fractures in the thoraco-lumbar region of the spinal column are relatively rare in automobile accidents.)

Other dynamic response criteria--Three computer simulation methods of particular note have been used for predicting thoraco-lumbar spine fracture injuries. The first two methods are related in that the second was developed as an extension of the first. The first method calculates a Dynamic Response Index (DRI). The second method--much more recently developed--is called the Acceleration Exposure Limit Method; it calculates an an "injury-risk criterion." The third method that is discussed below is a three-dimensional, discrete-element, head-spine model that predicts intervertebral stresses, which are used to calculate an Injury Potential Function.

The *Dynamic Response Index Method* (or Spinal Injury Model) is described in 1971 and 1975 references reviewed in the current study. Those references are by Brinkley and Shaffer (1971) and Payne (1975). The general method was first described by Payne (1962) and the DRI method specifically is introduced and discussed thoroughly in Stech and Payne (1969). The *Acceleration Exposure Limit Method* is described in reviewed 1988 and 1989 references: von Gierke et al. (1988) and Brinkley et al. (1989). Both models make use of a simple mass-spring-damper system for predicting gross response of an aircrew member in a system subjected to short duration acceleration loadings. Injury prediction by both models is calibrated by observational injury-level and injury-threshold data from various sources. A primary difference between the models is that the first, the DRI Method, considers +Z inputs to a one degree-of-freedom model, while the second, the Acceleration Exposure Limit Method, considers inputs and responses in three degrees of freedom, X, Y, and Z.

The DRI model determines the Z-response of a simple mass-spring-damper representation of the seated human. It has been used in relation to ejections and helicopter crashes. The DRI is the square of the natural frequency of the system (i.e., k/m, the spinal stiffness k divided by the head-plus-torso mass) multiplied by the maximum compressive deflection that results from a +Z driving force or acceleration in the simulation and divided by the acceleration of gravity:

$$\mathrm{DRI} = \frac{k/m}{g}\,\delta_{\max} = \frac{4\,\pi^2\,f^2}{g}\,\delta_{\max}\,.$$

The DRI is thus nondimensional. Brinkley and Shaffer reference system constants determined by Stech and Payne (1969) from experimental data--specifically, 0.224 for the damping ratio and 52.9 rad/s for the natural frequency, $2\pi f$. (The mass m and stiffness k do not occur separately in the equation of motion, but only as the ratio k/m.) Some of the injury calibration data were calculated from tests with cadavers and some are from operational experience (Payne, 1975). The spinal injury rate as a function of DRI is presented in (only approximate) semilogarithmic form by Payne (1975). The graph of Payne, from Brinkley and von Gierke (1973), is included here as Figure 3. (Also see Brinkley and Shaffer, 1971.) The results in Table 3, below, may be read from the Figure 3 graph (described as "preliminary" by Payne) for spinal fracture rate as a function of DRI.

Table 3. DRI vs. SPINAL FRACTURE RATE

DRI	Spinal Fracture Rate (%)
13.3	0.2
14.9	1.0
16.8	5.0
19.4	20.0
21.3	50.0

The more recently developed technique called the *Acceleration Exposure Limit Method* was introduced by von Gierke et al. (1988) and is described also by Brinkley et al. (1989). This method predicts the probability of injury due to combined, but independent, accelerations in X, Y, and Z axes. Therefore, while the DRI Method is suitable only for study of injury potential for +Z inputs, such as in ejections or some helicopter crashes, the Acceleration Exposure Limit Method has validity also in crashes with large fore-aft and lateral accelerations. Determined probabilities are based on acceleration limit values for specific levels of risk of injury. The acceleration limit values, for independent plus and minus X, Y, and Z accelerations, are derived from human impact data bases. In use of the model accelerations are presumed to have their greatest deleterious effect when

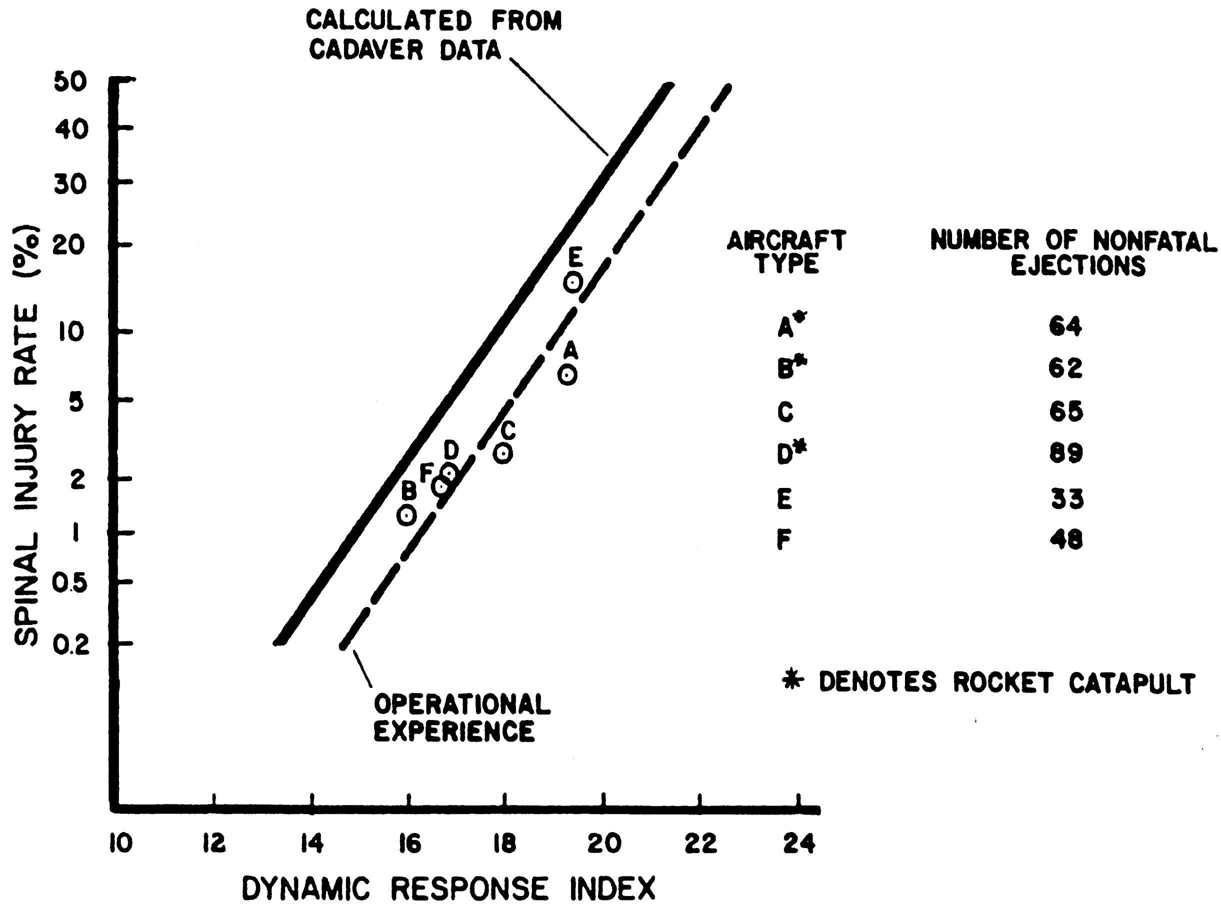


Figure 3. Probability of spinal injury estimated from laboratory data compared to operational experiences (from Payne, 1975, after Brinkley and von Gierke, 1973)

acting at a specific "critical point." That point is normally assumed to be the center of mass of the upper torso. Injury probabilities are estimated from the computed accelerations of that point.

Since aircrew member responses in X, Y, and Z in this model are assumed to be independent, the dynamic response accelerations have exactly the same form as the Dynamic Response Index for +Z in the DRI Model, i.e.,

$$DR_j = \frac{(k/m)_j}{g}\,\delta_{j,max} = \frac{4\,\pi^2\,f_j^{\,2}}{g}\,\delta_{j,max}\ ,$$

but each axis has a different natural frequency and maximum deflection. In the above equation the subscript "j" represents X, Y, and Z. Independence of X, Y, and Z responses results from the mass at the critical point being attached independently by three axial, spring-damper systems to the aircraft--or, in the case of ejection studies, to the ejection seat. The ejection seat, or whatever part of the aircraft is attached to the critical point, is assigned three linear acceleration components and an angular velocity. Nondimensional dynamic responses of the critical point mass are calculated by dividing the X, Y, and Z accelerations, DR_j, by the previously described acceleration limit values. A time-varying injury-risk measure is calculated as the square root of the sum of the squares of the three nondimensional accelerations. Thus, where ß is the injury-risk criterion, DR_x, DR_y, and DR_z are the dynamic response accelerations for the X, Y, and Z axes, and $DR_{x,L}$, $DR_{y,L}$, and $DR_{z,L}$ are limit values for each axis,

$$\text{ß} = \left[\,(DR_x\,/\,DR_{x,L})^2 + (DR_y\,/\,DR_{y,L})^2 + (DR_z\,/\,DR_{z,L})^2\,\right]^{1/2}.$$

Separate values of this measure are calculated for low-, moderate-, and high-risk limit accelerations. The escape system occupant is considered to have exceeded a specified injury-risk level if this injury-risk criterion has a magnitude greater than one. The limit acceleration Gs used by Brinkley et al., which were the best available data at the time of the study (1989), are given in Table 4 below. (The limit values for the -Z vector were determined by Brinkley et al. as a part of their reported research.)

Table 4. ACCELERATION LIMIT VALUES $DR_{j,L}$ FOR THE ACCELERATION EXPOSURE LIMIT METHOD (j=X,Y,Z)

Low Risk Limit Accelerations (Gs)		
+X = 35	-X = -28	
+Y = 14	-Y = -14 (w/o side panels)	±Y = ±15 with side panels
+Z = 15.2	-Z = -13.4	
Moderate Risk Limit Accelerations (Gs)		
+X = 40	-X = -35	
+Y = 17	-Y = -17 (w/o side panels)	±Y = ±20 with side panels
+Z = 18.0	-Z = -16.5	
High Risk Limit Accelerations (Gs)		
+X = 46	-X = -46	
+Y = 22	-Y = -22 (w/o side panels)	±Y = ±30 with side panels
+Z = 22.8	-Z = -20.4	

The recent references pertinent to the Acceleration Exposure Limit Method that were found in the literature search of the current study indicate that this method is still under evaluation.

A computer simulation model variously called the *Head-Spine Model (HSM)* and *SAM* (for the Structural Analysis of Man) is described in its first form by Belytschko and Privitzer (1978). The model is described further by Williams and Belytschko (1981), Privitzer et al. (1982), Belytschko et al. (1985), von Gierke et al. (1988), and Privitzer and Kaleps (1989). This model, which will be called HSM here, is a three-dimensional, discrete element model used for prediction of the dynamic response of the head-spine-torso structure to severe impact environments. It includes representation of the head, torso, pelvis, inter-vertebral discs, ligaments, muscle, and other connective tissues. The effects of muscle can be simulated with either a passive muscle model or a stretch reflex model. HSM is described as incorporating a data base that contains biomechanical, geometric, and structural data (Belytschko et al. 1985).

Privitzer et al. (1982) describe estimation of probabilities of fracture injury at separate levels of the spine from T1 to L5 by use of an injury criterion calculated by the Head-Spine Model, called the HSM Injury Function. This quantity represents the ratio, at each level of the spine, of the peak computed cortical shell compressive stress (due to combined axial compression and bending) to the ultimate compressive yield stress. The report does not give values for the ultimate yield stresses or a detailed definition of the HSM Injury Function.

Von Gierke et al. (1988) discuss an Injury Potential Function, which has a different value at each vertebral level and is obtained by dividing the maximum predicted stress at each level by the corresponding vertebral level mean failure stress. The Injury Potential Function is apparently the same as, or a refinement of, the HSM Injury Function referenced by Privitzer et al. (1982). Von Gierke et al. state that the Injury Potential Function has predicted the observed result of "higher probability of injury...in the middle thoracic region of the spine than in the lumbar region" in the case of "very tight torso restraint." Injury potential (probability) as determined from the Head-Spine Model is graphed in the paper as a function of vertebral level for four ejection simulations with peak $+G_z$s equal to 14, 16, 18, and 20 G. Von Gierke et al. indicate that an Injury Potential Function value of 1.0 for any particular vertebra indicates a 50 percent probability of fracture while a value of 0.9 indicates a 16 percent probability of fracture.

In a 1989 paper Privitzer and Kaleps describe a Spinal Injury Function, SIF, calculated by the Head-Spine Model. The SIF makes use of experimental compressive failure data of human thoraco-lumbar vertebrae, to predict the probability of injury, by level, along the thoraco-lumbar spine. The SIF is presumably a refinement of the HSM Injury Function described earlier by Privitzer et al. (1982). A Neck Injury Parameter, NIP, is defined in like manner. SIF and NIP values of 1.0 at any vertebral level correspond to a 50 percent likelihood of vertebral body compressive failure due to combined axial compression and bending at that level. The authors state that the injury prediction capability of the model has been validated using operational ejection data, but the validation work is not described in the paper. The paper does not give values for the ultimate yield stresses or detailed definitions of SIF or NIP or the corresponding injury criteria.

5.1.1.2 Ejection seat dynamics criteria. While the injury prediction methods discussed above in Section 5.1.1.1 are detailed in that they examine injury probability on a level-by-level basis along the thoraco-lumbar spine and/or include computer simulation techniques, another injury prediction method considers only the gross measures of ejectee response or the gross dynamic performance specifications of the ejection catapult, together with observational injury data. Those observational data are discussed in this section. Injury considerations in the literature that is relevant to gross dynamics of ejection systems almost invariably relate to thoraco-lumbar spinal fractures.

In theory the detailed methods of the former type have the greater potential for studying injury mechanisms; in practice, however, they place great demand on proper design of test manikins and discrete-element simulation programs and on proper interpretation of experimental and simulation results. Nonetheless, it may be the case that only such methods as those will be found adequate for refining design of ejection systems. The whole-body, ejection-dynamics criteria discussed in this section were, for the most part, determined in pre-1980 research focused on establishing appropriate limit values for gross dynamic performance characteristics of ejection systems. The DRI Method and Acceleration Exposure Limit Method of the preceding section are related to the whole-body, ejection dynamics discussed here, but since those methods--particularly the Acceleration Exposure Limit Method--make use of a great deal of experimental tolerance-to-acceleration data, they should have continued usefulness for directing and assessing development and refinement of escape systems.

Various parameters of the gross dynamics of the catapult and ejectee are discussed in the literature. These include peak $+G_z$ acceleration, the rate of onset of the acceleration profile, and velocity at end of stroke. Limit values and injury criteria estimated by various investigators are given below, but it must be noted that nearly all data of this sort in the literature assume a properly postured, properly restrained ejectee. It has been found by many researchers, as described in Section 4.2.2 and elsewhere in Section 4, that an erect, head-back posture with good torso and hip restraint is critical in reducing the rate of thoraco-lumbar fracture injuries for any ejection system.

A relatively recent reference (Naval Safety Center, 1981) does note specifically that for ejections in which the spinal column is properly aligned, an acceleration of +25 G_z can be supported without vertebral fractures. This same reference indicates that short duration accelerations from "seat slap" may be 40 G or more in through-the-canopy ejections without concomitant injury. Rates of onset of $+G_z$ acceleration as large as 500 G/s or more can be tolerated without injury if the ejectee is properly restrained and sitting erectly on a rigid, stable seat, according to this reference.

Nuttall (1971) summarizes human tolerances to short-duration, large-acceleration environments in terms of approximate values or ranges as follows: $+G_z$, 20 G; $-G_z$ (for downward ejection seat), 12 G; 250 G/s rate of onset, upward; 125 G/s rate of onset, downward; other values, $+G_z$ of 25 G and rate of onset of 300 G/s. The author notes that accelerations to the required ejection velocity should be over at least 230 ms. He makes reference to accidental noninjury-producing exposures of human subjects to 30-33 $+G_z$ at 500 G/s rate of onset in upward ejection experiments under ideal laboratory conditions.

There is more agreement in the literature on values for maximum supportable $+G_z$ acceleration and rate of onset than for duration of acceleration (or, almost equivalently, end-of-stroke velocity). Shannon (1971) cites 25 G as a conservative maximum limit for $+G_z$ and 500 G/s for maximum rate of onset but gives a range of 100 to 150 ms for duration. Discrepancies in the literature between cited duration values may be because some authors mean to indicate the maximum supportable duration for a given acceleration while others mean to indicate the minimum acceptable duration for accelerating the ejectee to the required ejection velocity.

In early ejection seat testing conducted with volunteer subjects, Watts et al. (1947a) find that 18 to 21 G was tolerated repeatedly without injury, but the authors do not reach a conclusion as to maximum $+G_z$ that can be tolerated under operational conditions. In a second report on their study Watts et al. (1947b) state that they believe 20 to 22 G to be the "practical upper limit" for seat ejection experiments with living human subjects. Catapult acceleration pulse durations were about 300 ms, strokes were 40 to 60 inches, and end-of-stroke velocities were up to about 60 ft/s. Maximum rates of onset for acceleration pulses were 150 to 280 G/s. Watts et al. (1947a) note that German researchers concluded in early work that fractures in the lumbar region will not occur until accelerations reach 22 to 25 G.

Table 5 below summarizes noninjury producing, limit values identified in the literature that was reviewed in the current study. It should be noted that it is not generally possible to use an ejection system that is designed with the most extreme values for all gross dynamics parameters; in general, tradeoffs are necessary.

Table 5. SUMMARY OF EJECTION SYSTEM DYNAMICS LIMITS FOR LOW RISK OF INJURY

Maximum $+G_z$	Maximum Rate of Onset	Minimum Pulse Duration	Maximum Change of Velocity
20-25 G	200-500 G/s	100-230 ms	20-60 ft/s

5.1.2 Cervical spine fractures. Fractures of C2, C5, and C6 are the most common major injuries to the neck that occur in ejections before complete egress from the aircraft. Nearly all existing injury criteria for the neck have come from automotive safety-related research even though neck injuries, except for strains, are relatively uncommon in automobile crashes. The mechanisms for vertebral fracture in the neck are complex, depending not only on the loads on and ultimate strengths of the vertebrae, but also very sensitively on initial positions and the conditions of loading. For the most part available data will not be of significant use in ejection system testing with manikins unless the neck of the manikin models the human neck in sufficient detail. The adequacy of the manikin neck can be established only by comparison of results from tests with manikins and cadavers or, indirectly, by confirmation of proper manikin prediction of ejection-related injuries seen in operational conditions.

5.1.2.1 Moment, force, and dynamic response criteria. The existing biomechanical injury tolerance data relevant to ejection-related neck injuries are of two types: (1) bending moment criteria; (2) neck force criteria, primarily for axial compression. Bending moment criteria are widely cited in the literature, but it is important to note here that nearly all values referenced originate from one particular study (Mertz and Patrick, 1972). A larger body of independent research relevant to axial compression injuries of the neck has been reported in the literature.

An observation by Patrick (1987) has possible importance to estimation of marginal injury level tolerances for both neck moments and neck forces (shear and axial). Patrick observes that, for the neck, cadaveric marginal injury level tolerances are about double the human subject maximum *voluntary* levels. The Mertz-Patrick data in Tables 6 and 7 below are consistent with this. Gracovetsky et al. (1982) and Helleur et al. (1984) also consider it reasonable to estimate injury level as a voluntary tolerance level multiplied by a constant. They address the question of whole-body acceleration levels rather than forces or moments in their papers, but they reference their earlier work that established that weightlifters will not voluntarily execute a lift that produces lumbar compression forces greater than two-thirds of ultimate strengths. Their findings would indicate that Patrick's hypothesis is conservative (i.e., that a factor of one and a half would be more appropriate than two) except that the weightlifters may have been more motivated to perform maximally than Mertz and Patrick's volunteer subjects.

Moment criteria--The study from which nearly all cervical moment-injury criteria cited in the literature derive was conducted by Mertz and Patrick (1972). Patrick also summarizes the results of the study in his 1987 paper. Human volunteers were subjected to dynamic environments that produced noninjurious neck responses in extension and flexion. Tests with cadavers were used to extend the data into the injury region. None of the tests involved direct impact to the head. Moments and forces at the occipital condyles were calculated from rigid-body motion equations by measuring head accelerations and estimating the inertial and geometrical characteristics of the head. Moment, shear force, and axial compression force injury criteria are given in the paper. Torque-deflection loading curves given are for angulation of the head with respect to the torso. Loading-unloading curve envelopes are defined for both flexion and extension. The response envelopes and some of the associated tolerance limits and injury levels determined by the authors are shown here in Figures 4 and 5 (Figures 26 and 28 of Mertz and Patrick, 1972). The moment-related tolerance levels for dynamic response determined by Mertz and Patrick are summarized here in Table 6 (from Patrick, 1987).

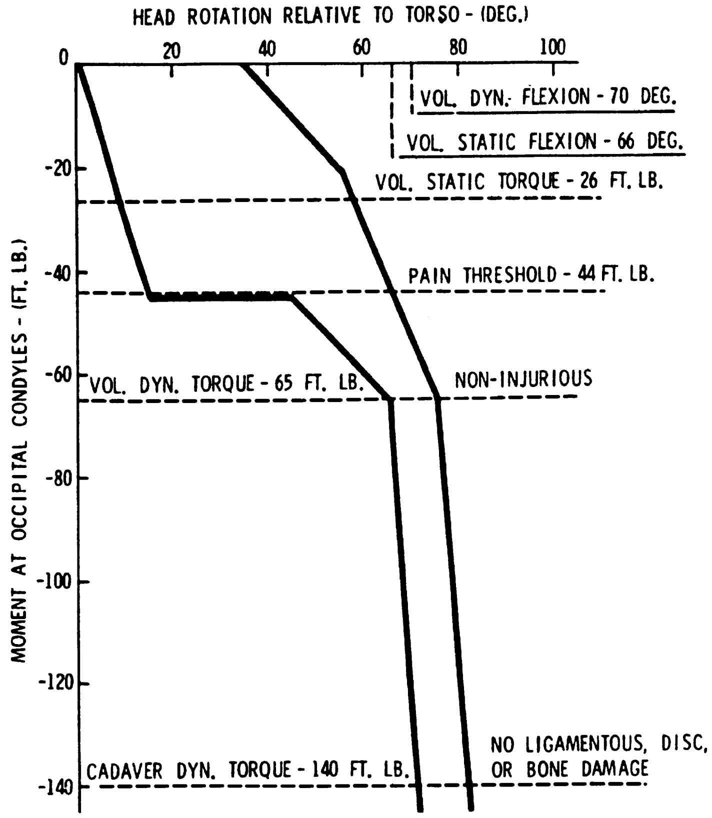


Figure 4. Head-neck response envelope for flexion and various tolerance levels (from Mertz and Patrick, 1972)

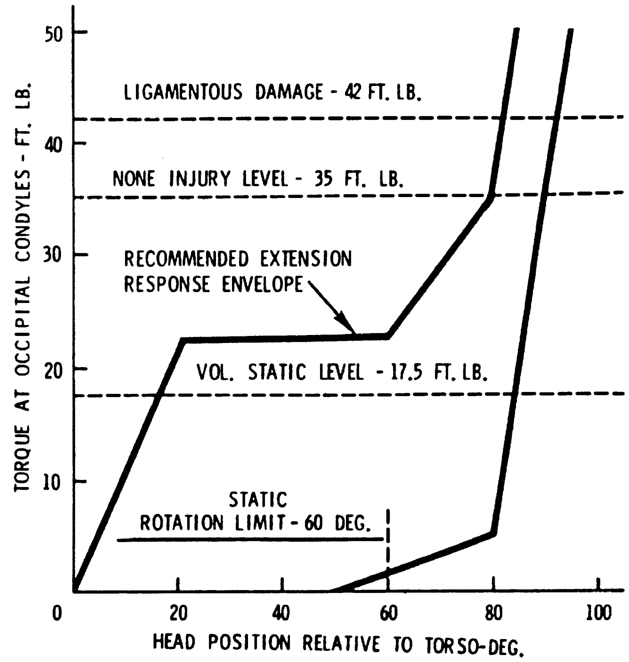


Figure 5. Head-neck response envelope for extension and various tolerance levels (from Mertz and Patrick, 1972)

Table 6. MERTZ-PATRICK NECK MOMENT TOLERANCES		
VOLUNTARY DYNAMIC MOMENT TOLERANCES AT THE OCCIPITAL CONDYLES		
Forward flexion (no injury)	65 ft-lb	(88 N-m)
Extension (no injury)	22.5 ft-lb	(30.5 N-m)
Lateral flexion (no injury)	33.3 ft-lb	(45 N-m)
CADAVERIC MARGINAL INJURY LEVEL TOLERANCES AT THE OCCIPITAL CONDYLES		
Forward flexion (no damage)	140 ft-lb	(190 N-m)
Extension (no damage)	35 ft-lb	(47.5 N-m)
Extension (damage to ligaments)	42 ft-lb	(57 N-m)

Mertz and Patrick, 1972; Patrick, 1987

The injury criteria established by Mertz and Patrick are conservative in that cervical fracture did not occur in any of the cadaver (or volunteer) tests. The most severe injury that occurred was ligament damage in cadavers. The mistake should not be made of assuming no significant injury to living human beings at moment loadings that did not produce ligament or vertebral injury in cadavers, since severe strains and neurological damage can surely occur. Nonetheless, it may be true that the injury criteria of Mertz and Patrick do not have great relevance in studies of neck injury resulting from aircrew member ejections. Studies of neck injury in automobile accidents have consistently indicated that cervical fractures are rare in the absence of head impact (e.g., Portnoy et al. 1972; Cheng, 1982; Ommaya, 1984). It is not clear from the literature review of the current study that this question has ever been addressed directly in studies of ejection-related cervical fractures.

Force criteria--Mertz and Patrick also determined voluntary static tolerance levels for shear and axial forces in the neck. They report only one dynamic force tolerance value (cadaveric, anterior-posterior shear force). Their force tolerance levels are summarized in Table 7.

Mertz and Patrick state that the voluntary static force tolerances determined in their study (given in the above table) can be considered lower bounds for marginal injury level forces. This is certainly true, but a number of studies since the Mertz-Patrick study (1971-1972) have determined the actual minimum *fracture*-producing axial compressive loads to be on the order of 1000 lb or greater--i.e., much larger than the voluntary tolerance of 250 lb. Melvin (1979), for example, states that fracture of cervical vertebrae occurs for compression loads of

Table 7. MERTZ-PATRICK NECK FORCE TOLERANCES

VOLUNTARY STATIC FORCE TOLERANCES AT THE OCCIPITAL CONDYLES

A-P or P-A shear force	190 lb	(845 N)
L-R or R-L shear force	90 lb	(400 N)
*Axial compression force	250 lb	(1110 N)
*Axial tension force	255 lb	(1135 N)

CADAVERIC SUB-INJURY LEVEL RESPONSES AT THE OCCIPITAL CONDYLES

A-P shear force (no damage)	450 lb	(2000 N)

*Mertz and Patrick, 1972; Patrick, 1987

about 1280 lb. McElhaney et al. (1983) find that in dynamic compression loading of the full cervical spine burst fractures of the C5 vertebral body are common and require 1400-1800 lb. Other experimental studies that suggest large values for ultimate strengths of cervical vertebrae are discussed below and include Culver et al. (1978) and Cheng et al. (1982).

On the otherhand, there are also reported experimental results that are more in accord with the suggestion by Mertz and Patrick that 250 lb be used as a lower bound for marginal injury level forces. Those studies are also discussed below. They include Pintar et al. (1989) and Hodgson and Thomas (1980; 1983). Additionally, to the degree that it is valid to extrapolate upper-thoracic vertebral strength data to the cervical spine, the strength-versus-level data previously presented in equation form for vertebral levels T1 and below are suggestive that low values are appropriate (see Table 2 in Section 5.1.1.1). For example, for C5, i.e., L=-2, it may be appropriate to extrapolate the T1-L5 data of Coltman et al., "AVERAGE", "GREATEST", and "LEAST BOUND"; the T1-L5 data of Kazarian and von Gierke, "fast loading" and "slow loading"; the C4-L1 data of Payne (Messerer); and the T1-T5 data of Henzel. The respective results, for C5, are: 20 lb, 689 lb, -25 lb; 259 lb, 398 lb; 706 lb; and 0 lb.

It is apparent that cervical vertebra compressive strengths have low values, typically less than 500 lb, under conditions of quasistatic loading, and that strengths are larger--greater than 1000 lb--for short-duration, dynamic loading. Experimental dynamic loadings are usually accomplished by crown (top-of-head) impacts by padded impactors of mass 10 kg and impact velocities of 2 to 11 m/s. (Isolated cervical spine preparations are sometimes loaded dynamically as well.) Since the conditions of dynamic loading experiments are much more like manikin or aircrew member ejections than are quasistatic loadings, it is probably appropriate to use the larger ultimate strength data (or the Hodgson-Thomas criteria) in interpreting manikin test data.

Citing data from a study by Culver et al. (1978), the SAE Information Report SAE J885 JUL86 (Society of Automotive Engineers, Inc., 1991b) indicates that in cadaver crown-impact tests peak loads of less than 1560 lb usually did not produce neck fractures. In eleven tests the lowest load that produced vertebral process fractures was 1060 lb. That test also produced some lateral lip crush of the C5 body. The lowest load that produced significant crush of any vertebral body (C5) was 1620 lb. A test with an superior-inferior (S-I) head load of 1990 lb crushed the C3-4, C4-5, and C5-6 discs, fractured two transverse processes, and severely crushed the T2 vertebral body. The authors comment that since the measured forces were dynamic head loads (for a padded impactor), the corresponding axial compressive neck loads could be smaller due to head mass inertial effects. Additional note is made that fractures can occur at on the order of half these loads if the head, neck, and torso are not axially aligned. With regard to shear force injuries in the neck, the authors note that the upper part of the neck (occipital condyles to C2) is most subject to injury. An implication is that it is important to measure shear force in the upper neck in manikin tests. This SAE Information Report does not, however, give injury criteria data for the neck in shear.

A study is reported by Pintar et al. (1989) in which quasi-static, compressive loading tests of seven fresh human cadaveric head-neck complexes were conducted. Six-axis load cells were placed at the proximal and distal ends of the specimens to document the gross biomechanical response. The preparations were loaded axially to failure at a rate of 2 mm/s. At failure the preparations were deep frozen in the compressed state to preserve tissue alterations. Failure loads ranged from 1355 N to 3612 N (305 lb to 812 lb) for the seven preparations, while deflections at failure ranged from 9 to 37 mm. Strains at failure ranged from approximately 0.04 to 0.26 mm/mm. Upper cervical injuries were observed under compression-extension modes while lower cervical injuries occurred under compression-flexion modes.

Dynamic impact loading of the neck through direct head impacts of cadaveric subjects was studied by Hodgson and Thomas (1980; 1983) with regard to numerous variables. These included impact location, line of action, energy level, concentrated versus distributed loading, initial neck curvature, and protective gear. Among the most important findings is that, for compression loading in general, there are too many variables affecting cervical spine injury to publish suggested tolerance limits. Nonetheless, the authors do present their "best estimate of axial compression tolerance for the adult population." An aspect of their neck-injury criterion that is different from any others found in the literature is a dependence on duration of loading over a given force level. (This feature is seen in some head and chest injury criteria.) Specifically, for the adult population, they estimate potential for serious injury for an axial compression force of 250 pounds or more for a duration of 30 ms or more or for force greater than 850 - 20 x T pounds for T less than or equal to 30 ms, and no injury otherwise. No statement is made regarding an anticipated injury site. (The authors give an upper bound criterion as follows: potential for serious injury for an axial compression force of 250 pounds or more for a duration of 36 ms or more or for force greater than 1450 - 33.3 x T pounds for T less than or equal to 36 ms.)

In consideration of this qualified injury criterion presented by Hodgson and Thomas, Eppinger (1982) chooses to make a more conservative interpretation. Where Hodgson and Thomas say (1983, page 115, Figure 5) that the injury criterion indicates a "potential for serious injury," Eppinger says that it should indicate an AIS 5 injury (critical).[1] Eppinger explains (1982, fifth unnumbered page) in this way: "Because neck injuries are either minor or catastrophic, it is difficult to apply a continuous scaling of AIS versus some mechanical input. Before C1 and C2 separate no serious injury is likely. Once they do separate, death is assured."

With regard to compressive-loading neck injuries that might be caused by ejection forces, the results presented by Hodgson and Thomas (1980; 1983) are consistent with a conclusion that cervical compression injuries--like thoraco-lumbar compression injuries--are more likely when the neck is flexed, that is, when the ejectee is not seated erectly with head and buttocks back against the seat. (The authors do not discuss ejections specifically.) Their results also support the view that the site of injury (C1 to C7) is greatly dependent on line of action of ejection acceleration and initial head-neck orientation.

Cheng et al. (1982) conducted a study that may not be of great pertinence to ejection injury research since primary accelerations were in $-G_x$ rather than $+G_z$, but some information in their paper is of interest. Six cadavers were tested in chest impacts of severity great enough to produce cervical fractures and fracture dislocations without head impact. On the basis of experimental results for a combined axial tension and flexion mode of inertial loading of the neck, the authors propose a neck fracture criterion of 1400 lb resultant neck load (vector sum of axial tension and shear forces) at the base of the skull. The proposed value is described as a "conservative" indicator of probable fracture at the atlanto-occipital joint. It is stated that in the combined axial tension and flexion mode, the critical parameter governing injury is axial tension and that the role of shear and moment is unclear.

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[1]AIS is the Abbreviated Injury Scale (AIS) severity code, which is defined for all types of injury for all parts of the body in the 1990 reference from the Association for the Advancement of Automotive Medicine. AIS is assigned for each injury on a 0-6 ordinal scale. The AIS values correspond to general levels of injury as follows: 0=none; 1=minor; 2=moderate; 3=serious; 4=severe; 5=critical; 6=maximum.

Finally, it should be stressed that adoption of *conservative neck-injury criteria* for trauma assessment in ejection studies is important. The great sensitivities of injury probability to factors such as initial neck curvature, initial orientation of the ejectee's thoraco-lumbar and cervical spines to the acceleration vector, and effectiveness of upper torso restraints have already been discussed in relation to observations of ejection outcomes and various experimental studies. Section 5.1.2.3 discusses some related factors further. Experimental prediction of such sensitivity, and associated injuries, using test manikins is very difficult and fraught with potential for error both in experimental procedures and in interpretation of results. For these reasons alone, use of conservative neck-injury criteria is dictated, but there are additional reasons, as well. One is that there is evidence that serious neck injuries can occur with loadings less than loads normally thought necessary to produce such injuries. Schall (1989), for example, documents eight non-ejection cervical spine injuries, including vertebral fractures, of aircrew members of F-15 or F-16 aircrew. All of the injuries are attributed to $+G_z$ forces during high performance maneuvers. They include two compression fractures, three left herniated nucleus pulposus, one fracture of a spinous process, one interspinous ligament tear, and one myofascial syndrome. An additional, and important, reason for use of conservative neck injury criteria is the criticality of some neck injuries. Yoganandan et al. (1989a) discuss auto accident-related spinal injury data that surely should be considered in the use of manikin test data and in the design of escape systems. Their paper examines a large amount of crash victim clinical data and accident data base information. Although the distribution of spinal injury types is different from that for aircrew member ejections, it is clear from the findings that emphasis should be placed in manikin studies on reduction of flexion-compression loadings of the cervical spine and shear loadings at the craniocervical junction. The former is responsible for nearly all cases of complete and incomplete quadriplegia in auto accidents and the latter is responsible for nearly all spinal-column related deaths.

There is additional discussion in Section 5.1.2.3 (<u>Head impact force factors</u>), below, that has pertinence to force-related criteria for neck injury. That section describes non-quantitative considerations important in assessment of the relationship of head-canopy forces to neck injury in through-the-canopy ejections, but also important to some degree in understanding mechanisms of neck injury in the absence of head contacts.

<u>Other dynamic response criteria</u>--It can probably be properly assumed that neck injuries that occur in aircrew member ejections do not result from noncontact, large-angle motions of the head relative to the torso. Further, axial compression forces in the neck almost surely have bearing on ejection-related neck injuries that occur prior to complete egress from the aircraft. Because

of these factors, recent research results reported by Kallieris et al. (1991) will likely not be relevant to the study of neck injuries in manikin ejection tests. Their work is discussed briefly here, however, because of its possible relevance and because it is research of a sort that has not previously been seen in the literature. Twenty-three frontal ($-G_x$) car crashes with a vehicle crash barrier and fourteen car-to-car lateral ($+G_y$) collisions were conducted. Cadaver subjects were used, and they were restrained by three-point belt systems in frontal crashes and by belts and a door panel in lateral crashes. Thus, acceleration inputs to the torso of the cadaver were primarily either $-G_x$ or $+G_y$ rather than $+G_z$ as in ejections. High-speed film analysis and accelerometer data determined head and neck angular and translational displacements, velocities, and accelerations. At sufficiently high crash accelerations neck injuries occurred in the absence of head contacts, i.e., injuries could be attributed to forward and lateral flexion of the head/neck. The highest correlation of any determined response with severity of neck injury was maximum head translational acceleration in the direction of the trajectory (i.e., along the path). Above a value of 21 G for this response, there was injury in every case. Compression fracture of the cervical vertebrae was uncommon. Rupture of the intervertebral disc was the most common of all types of cervical injuries observed.

5.1.2.2 Ejection seat dynamics criteria. Only a single reference reviewed in the current study relates established limits on the gross dynamics parameters of ejection systems to neck injuries seen in operational ejections. Instead, in other references the established limits, such as a 20-25 G limit for $+G_z$, are consistently related to thoraco-lumbar fracture injury. Section 5.1.1.2 discusses and summarizes the pertinent literature reviewed in this study as regards thoraco-lumbar injury.

The one such study that has pertinence to the neck is a simulation study with a mathematical model of the upper spine (C1 to T6) and skull. Gracovetsky et al. (1982) determine a 40 G maximum "supportable" (noninjurious) acceleration for best-case neck posture and orientation with respect to the $+G_z$ acceleration vector. For worst-case neck posture and orientation they determine a value of 13 G. Their results are based on the simulation values of vertebral compressive stresses.

5.1.2.3 Head impact force factors. Available data suggest that head-canopy forces in through-the-canopy ejections may be responsible for a greater incidence of neck fractures than seen for jettisoned-canopy ejections. (See Section 4.1.1.) While the current study has reviewed only one experimental research report that examines head-canopy forces in through-the-canopy ejections (Chiou et al. 1993; see Section 4.2.4), the literature review does include papers from a body of recent automotive safety-related research with cadavers in which neck injury results from crown impacts. Some of the findings from those studies are described here.

Nightingale et al. (1991) measured the passive combined flexion and axial loading responses of the unembalmed human cervical spine. They found that different end conditions (unconstrained, rotational constraint, and full constraint) greatly influenced the risk of injury, the failure mode, and the observed axial load to failure. These general findings by Nightingale are in full accord with the findings of other researchers who have studied impact and quasistatic loadings of the head in cadaveric head-neck or whole-body specimens. The implications to research involving head impacts of manikins are important--namely, that the neck module of the manikin must respond like a human neck for a variety of loading modes if detail regarding neck-injury mechanisms is to be derived from manikin studies and, further, that manikin neck response data must be interpreted with great care and consideration for differences between manikin and human necks. The results of the study by Nightingale et al. suggest that safety equipment and injury environments should be designed to minimize the degree of imposed constraint on the head. In particular, systems that tend to "pocket" may produce an enhanced injury potential. This finding is consistent with a finding in a simulation study by Bowman and Schneider (1980) (also Bowman et al. 1981) that lessening the coefficient of friction between a helmet and a struck surface, particularly for crown impacts, can significantly reduce the likelihood of neck injuries.

The axial load to failure for lower cervical bilateral dislocation was found to be significantly lower in the study by Nightingale et al. than the axial load to failure for vertebral compression-type fractures. The fact that cervical vertebral compression fractures often occur with absence of lower cervical bilateral dislocation is due to the great sensitivity of outcome to loading conditions and initial positions. Schall (1989) notes that cervical fractures can occur during flexion and extension at approximately half the axial load required to cause fracture in the absence of flexion or extension. Comment is made in the SAE Information Report SAE J885 JUL86 (Society of Automotive Engineers, Inc., 1991b) that nonalignment axially of the head, neck, and torso can reduce by half the neck compressive loads necessary to cause cervical fracture. These findings are in qualitative agreement with observation in ejection-injury studies.

McElhaney et al. (1988) studied the lateral, anterior, and posterior passive bending responses of the human cervical spine using unembalmed cervical spinal elements obtained from cadavers. Many of their tests were done with combined axial loading of the neck. Bending stiffness was measured in six modes including compression-flexion. Loads and moments at failure were also determined. End conditions were found to have a large effect on measured bending stiffness, with values being eight times as large for fixed-pinned conditions as for pinned-pinned conditions. McElhaney did not study impact loading of the head, but the maximum value of quasistatic, axial neck load was found by them to be a poor indicator of the type and magnitude of failure stresses.

Alem et al. (1984) report a study that investigated nineteen impacts to the head in the superior-inferior direction using unembalmed cadavers. Some impacts were used to study sub-injurious response and to determine mechanical characteristics of the system. The 10-kg impactor produced cervical spine injuries for impact velocities between 7 and 11 m/s. In agreement with McElhaney et al. (1988), these researchers determined that peak impact force is not a reliable predictor of cervical injury, nor is HIC (the Head Injury Criterion). Peak head linear velocity was the best indicator of injury of all response parameters measured. The maximum value for which there was no ligament, disc, or vertebra damage in the neck was 3.7 m/s, and the minimum for which damage did occur was 3.5 m/s. Of the impact parameters examined, the integral of the impact force-time curve (the impact impulse) was the most consistent indicator of cervical injury. The maximum value for which there was no neck injury was 36 N-s, and the minimum for which damage did occur was 35 N-s.

Huelke and Nusholtz (1985) describe experiments in which superior-inferior crown impacts were delivered to cadavers by either a guided moving impactor mass (56 kg) or a free-fall drop of the test subject. They found that peak impact force is not a good predictor of cervical injury and that flexion-type injuries are unlikely when the head and neck are constrained to move only in the midsagittal plane. They found also that the clinically described "head bowing to the chest" is not necessary for flexion-type injuries. Flexion-type cervical spine damage was observed in some cases with extension head motion and extension-type damage was observed with maximum flexion motion. The authors believe that many flexion-type injuries occur before gross head motion.

In a paper that discusses clinical neck injury data mostly from automobile accidents and involving head impacts of all types (not only crown impacts), Ommaya (1984) stresses the importance of minimizing the degree of head impacts since this reduces the potential for both head injury and neck injury. He notes that serious neck injury seldom occurs in the absence of head contact. This opinion is expressed by many other automotive safety researchers as well (e.g., Portnoy et al. 1972; Culver et al. 1982).

McElhaney et al. (1983) summarize findings from automobile and motorcycle accident injury studies with regard to causation of fractures of cervical vertebrae. They note that the most commonly seen fractures are of C1, C2, and C5--which is nearly the same as the C2-C5-C6 distribution seen in ejection-related cervical fractures. C1 and C2 fractures occur for low facial impact (extension-tension). C5 extension-compression injuries occur for high facial impacts, and C5 flexion-compression injuries occur in crown impacts. Fractures in the lower neck occur at C4 and C6 with about half the frequency of those at C5.

5.2 Biomechanical Properties. Although biomechanical properties--such as stiffness and damping characteristics--of the cervical and thoraco-lumbar spinal columns and their elements have been documented in the current study, that information is peripheral to the focus of the study. Accordingly, biomechanical properties will not be discussed here. The interested reader is directed to Table 9 for related information.

TABLES 8 AND 9

Summaries of references pertinent to phases of ejection escape preceding complete egress

Table 8 -- Injury Priority Analysis (Subtask 2)

Table 9 -- Trauma Assessment Criteria (Subtask 3)

Table 8. INJURY PRIORITY ANALYSIS (Subtask 2)
Summary of References

<table>
<tr><th colspan="3">REFERENCE</th><th colspan="4">ESCAPE SEQUENCE PHASES OF PERTINENCE</th><th colspan="4">PARAMETERS ADDRESSED</th><th colspan="7">INJURY TYPES ADDRESSED</th></tr>
<tr><th>Author(s)</th><th>Date</th><th>Title</th><th>Type of Aircraft Maneuver</th><th>Crewmember Pre-Escape Positioning</th><th>Ejection Boost Forces</th><th>Helmet Impact w. Canopy</th><th>Air-craft Speed</th><th>Severity of Maneuvers</th><th>Mission Require-Ments</th><th>Crewmember Physiology and Anthro-pometry</th><th>Cervical Spine Dis-locations/ Fractures</th><th>Thoracic, Lum-bar, & Sacral Dislocations and Fractures</th><th>Musculature & Ligamen-tous Sprains and Strains</th><th>Joint Disloca-tions</th><th>External and Internal Con-tusions and Bleeding</th><th>Concus-sions</th><th>Burns</th></tr>
<tr><td rowspan="2">Chiou, W.-Y.; Ho, B.-L.; Kellogg, D. L., Jr.</td><td rowspan="2">1993</td><td rowspan="2">Hazard potential of ejection with canopy fragmentation</td><td></td><td></td><td>X</td><td>elimin-ated</td><td>0 to 600 kn</td><td></td><td></td><td></td><td>compress-ive fx</td><td>DRI calculated</td><td></td><td></td><td>from frag-mentation</td><td></td><td>X</td></tr>
<tr><td colspan="15">SUMMARY / COMMENTS: Canopy fragmentation by small explosive cords immediately prior to ejection facilitates through-the-canopy ejection, which in itself has elimination of pre-ejection time delay as a primary advantage over jettisoned-canopy escape. This study examines the injury producing potential of fragmentation, high sound levels, and other factors in this alternative ejection method. The hazard potential of fragmentation was evaluated by a series of horizontal sled tests at sled speeds of 0, 150, 350, and 600 knots. The canopy was fragmented in nine of 14 through-the-canopy ejections with test dummies. It was found that canopy fragmentation reduced the mean compressive neck load from 231 kg to 108 kg. It was found that the likelihood of injury to aircrew members from sharp edges of fragments or from impact by pellets of the lead skin of the explosive cord is not significant. Similarly, there was found to be only very minimal associated risk of hearing damage. The conclusion was reached that no significant hazard is added by use of MDC's (miniature detonating cords) for canopy fragmentation and that the probability of spinal injury is still the main concern in escape ejections.</td></tr>
<tr><td rowspan="2">Hamalainen, O.; Vanharanta, ; H.</td><td rowspan="2">1992</td><td rowspan="2">Effect of Gz Forces and Head Movements on Cervical Erector Spinae Muscle Strain</td><td>combat, high-performance</td><td>head-neck</td><td>X</td><td></td><td></td><td>+7 Gz bank
+4 Gz LOOP
+4 Gz DU
+4 Gz MANE
-1.5 Gz NEGA</td><td></td><td>maximal voluntary contraction of cervical erector spinae muscles</td><td>X</td><td></td><td>cervical</td><td>cervical</td><td></td><td></td><td></td></tr>
<tr><td colspan="15">SUMMARY / COMMENTS: During flight missions the EMG activity of the cervical erector spinae muscles was measured for ten fighter pilots. Under +7.0 Gz the mean muscular strain was 5.9-fold compared with +1.0 Gz and was 37.9% of the maximal voluntary contraction (MVC). In some individuals the muscular tolerance (100% of MVC) was ipsilaterally reached already under +4.0 Gz with concomitant movements and twisted positions of the head. Pilots are susceptible to acute neck injury when the protection afforded by their neck muscles is insufficient. Pilots in the study experienced severe neck pain at +Gz's of much less than boost accelerations in ejection.</td></tr>
<tr><td rowspan="2">Visuri, T.; Aho, J.</td><td rowspan="2">1992</td><td rowspan="2">Injuries associated with the use of ejection seats in Finnish pilots</td><td></td><td>X</td><td>X</td><td></td><td>300 to 700 kph</td><td></td><td></td><td>X</td><td></td><td></td><td></td><td></td><td></td><td></td><td></td></tr>
<tr><td colspan="15">SUMMARY / COMMENTS: Injuries associated with 17 uses of ejection seats by Finnish pilots from 1958 to 1991 were analyzed. Three escapes were completed without any observed injuries. There was one fatality due to direct impact with a tree after low-altitude ejection. Of the remaining 13 five involved major injuries (requiring hospital treatment or longer sick leave) and eight involved lesser injuries. Of the five with major injuries three had compression fractures of a thoracic vertebra thought to be from ejection forces. One had a fracture of the femur caused by contact with the railing. The fifth had a rupture of a ligament of the right knee, which was caused by the leg restraint system. Minor injuries incurred in eight ejections included bruises, contusions, and strains. The effect of the height, weight, body/mass index, and the age of the pilots on the severity of the injuries was tested using a one-way analysis of variance. No statistically significant relationships were found.</td></tr>
<tr><td rowspan="2">Yacavone, D. W.; Bason, R.; Borowsky, M. S.</td><td rowspan="2">1992</td><td rowspan="2">Through the canopy glass: a comparison of injuries in naval aviation ejections through the canopy and after canopy jettison, 1977 to 1990</td><td></td><td></td><td></td><td>X</td><td></td><td></td><td></td><td></td><td>X</td><td></td><td></td><td></td><td></td><td>X</td><td></td></tr>
<tr><td colspan="15">SUMMARY / COMMENTS: The primary purpose of the study presented was to examine the comparative safety of two methods of ejection from tactical aircraft, viz., ejecting through a closed canopy and jettisoning the canopy prior to seat travel. The ejection data base of the Naval Safety Center was used, with the data search limited to the period 1977 through 1990. Only the 916 ejections in which injuries were coded by the reporting flight surgeon as "from ejection sequence" were considered. Minor injuries occurred with nearly the same likelihoods for the two methods of ejection, but through-the-canopy ejection produced more severe injuries with greater likelihood: 10.7% vs. 4.7% for fatalities and 29.2% vs. 17.4% for "at least one workday lost." The paper contains a small amount of information pertinent to ejection injury priority analysis. It is noted that fatalities were attributed to "multiple trauma" in about the same percentages, 77.7% and 70.3%, respectively, for the two escape systems; these injuries occurred during ejection but mostly post egress. For both about 20% (19.4% and 22.2%) of fatalities were attributed to skull-cervical fracture injuries. However, in the case of through-the-canopy ejections these are attributed mostly to striking the canopy while for jettisoned canopy cases they are attributed mostly to striking part of the aircraft post egress.</td></tr>
<tr><td rowspan="2">Crowley, J. S.</td><td rowspan="2">1990</td><td rowspan="2">Helicopter aircrew helmets and head injury: a protective effect</td><td>helicop-ter crash</td><td></td><td></td><td>helmeted head im-pact with interior</td><td></td><td></td><td></td><td></td><td></td><td></td><td></td><td></td><td>X</td><td>X</td><td></td></tr>
<tr><td colspan="15">SUMMARY / COMMENTS: Head injuries in helicopter crashes were studied. No estimates of head impact speeds are given. A strong inverse correlation between helmet use and the severity of head injuries is noted, but although other types of injury were studied, no note is made of an observed relationship between helmet use and the severity of neck or other types of injury. If, in fact, no relationships are present, and if helicopter vertical speeds at impact are similar to helmet/canopy impact speeds in through-the-canopy ejections, then this might be support for a view that the nature of helmet/canopy impact in ejections has no significant bearing on the nature of neck injuries that occur, if any. This means that other factors, such as catapult or rocket boost acceleration, are more important.</td></tr>
<tr><td rowspan="2">Freivalds, A.; McCauley, D.S.</td><td rowspan="2">1990</td><td rowspan="2">Biodynamic simulations of helmet mass and center-of-gravity effects</td><td></td><td>head and neck initial resting angles (simulation)</td><td>AV-8B ejection profile (simula-tion)</td><td>(none)</td><td></td><td></td><td></td><td></td><td>simulation measures only (neck)</td><td></td><td>simula-tion measures only (neck)</td><td>simula-tion measures only (neck)</td><td>simula-tion measures only (head)</td><td>simula-tion measures only (head)</td><td></td></tr>
<tr><td colspan="15">SUMMARY / COMMENTS: Ejections were simulated with a computer model with different helmet weights and centers of gravity and also different initial resting angles of the head and neck. It was found that added helmet mass has little effect on the likely severity of head or neck injury due to pulse impact acceleration. If, however, the helmet design or initial head position is such as to put the head-helmet center of gravity forward of the head/neck joint, the severity of head and neck injury is likely to be much greater for any normal ejection profile. [AV-8B ejection characteristics were used. The catapult accelerates the seat at a constant rate of acceleration from 0 G to 16 G in the first 80 ms. From 80 ms to 170 ms, the catapult maintains a constant acceleration of 16 G. At 170 ms, the rockets take over, adding an acceleration of 12 G at 45 degrees to the ejection angle (which is 19 degrees rearward).]</td></tr>
</table>

Table 8. INJURY PRIORITY ANALYSIS (Subtask 2)
Summary of References

REFERENCE			ESCAPE SEQUENCE PHASES OF PERTINENCE				PARAMETERS ADDRESSED				INJURY TYPES ADDRESSED						
Author(s)	Date	Title	Type of Aircraft Maneuver	Crewmember Pre-Escape Positioning	Ejection Boost Forces	Helmet Impact w. Canopy	Air-craft Speed	Severity of Maneuvers	Mission Require-Ments	Crewmember Physiology and Anthro-pometry	Cervical Spine Dis-locations/ Fractures	Thoracic, Lum-bar, & Sacral Dislocations and Fractures	Musculature & Ligamen-tous Sprains and Strains	Joint Disloca-tions	External and Internal Con-tusions and Bleeding	Concus-sions	Burns
Guill, F. C.; Herd, G. R.	1989	Aircrew neck injuries: a new, or an existing, mis-understood phenomenon?	X	X	X	X	X	X	X	X	X		cervical				
			SUMMARY / COMMENTS: The injury records for all U.S. Navy aircrew ejections from 1949 through March 1988, 4335 in total, were studied. The focus of the study was on "ejection associated" neck injuries. The authors reach several conclusions. One is that neck injury rates are increasing, although cervical fracture rates are increasing more slowly than paracervical sprains and strains. They are also of the opinion that there is no single, primary causal factor but that, rather, the underlying causal factors are many and varied. They make special note, however, that a significant proportion of the serious ejection associated neck injuries are likely to have been induced by the inflight maneuvering/gyration forces imposed upon the aircrew prior to ejection or during ejection. Inflight maneuvering becomes particularly important as helmet masses and moments of inertia are increased by additional instrumentation. Among factors examined and found not to be primary were aircraft maneuver/gyration forces, type of mishap, poor body position, escape system malfunctions, pre-existing cervical injuries, ejectee anthropometry, canopy mode, and ejection airspeed. Despite their finding that the rate of neck injuries is increasing, the authors find that neck injuries still constitute a small percentage of all ejection-related injuries. From 1949 to 1968, in 1764 ejections, among survivors there were a total of 12 cervical fractures and 112 cases of sprain or strain. From 1969 to 1988, in 1677 ejections, the corresponding numbers were 28 and 204.														
Guill, F. C.; Herd, G. R.	1989	An evaluation of proposed causal mechanisms for "ejection associated" neck injuries	X	cervical vertebral alignment	X	through-the-canopy vs. jettison canopy	0 to 500 kn	X	X	X	X		(cervical)	(cervical)			
			SUMMARY / COMMENTS: This paper analyzes moderate and severe neck injuries reported by U.S. Navy pilots after ejections. Data are from August 1949 through March 1985. The paper is comprehensive in its discussion of "ejection associated" neck injuries. It examines a complete range of types of neck injury and all conceivable potential causal factors/mechanisms related to all phases of the emergency escape, from pre-ejection through ground rescue. SUMMARY from paper: Dynamic response-type neck injuries are, in fact, experienced both by non-ejectees during flight and ejectees. There is strong anecdotal evidence which, when coupled with ejection report data, suggest that a significant proportion of those ejectees sustaining an "ejection associated" dynamic response-type neck injury might well have sustained their injury prior to the ejection, during the aircraft maneuvers/gyrations preceding the escape. There is evidence that many of the reported neck injuries were the consequence of system malfunction, i.e., seat striking ejectee during parachute opening following man-seat separation, entanglement of the ejectee with the seat prior to parachute opening. Several others among the reported injuries appear to have occurred during the parachute landing fall or during the rescue attempts. An important aspect of this problem is that the severest neck injuries--the fractures, subluxations, transections and herniations or rupture of intervertebral discs--occur relatively infrequently and seemingly at a constant rate. Neck injuries associated with ejections from aircraft do not conform to the patterns expected for any single proposed causal factor and mechanism which have been advanced to date. The data are too sparse to make simultaneous multiple partitions of the data to obtain better insight into the likely effects of the various proposed causal factors. The GENERAL FINDINGS include: the frequency of neck injuries among U.S. Navy ejectees has been gradually increasing over the past 15 years; severities have remained relatively constant during the past 15 years; the frequency of severe injuries of the neck region is significantly lower than the frequency for the non-neck region; severe neck injuries occur mostly at C2, C5, and C6.														
Guill, F. C.; Herd, G. R.	1989	Ascertaining the causal factors for "ejection-associated" injuries	detailed statistics	X	X	X	X				stats for assigned causes, 1949-1982	statistics for assigned causes 1949-1982	cervical and thoracic/ lumbar; detailed statistics				X
			SUMMARY / COMMENTS: This paper discusses injuries of all sorts reported by U.S. Navy pilots after ejections. Data are from 1949 through 1982. The paper is comprehensive in its discussion of "ejection associated" injuries. It examines injuries of various types and all conceivable potential causal factors/mechanisms related to all phases of the emergency escape, from pre-ejection through ground rescue. The paper says that determining the cause(s) for an ejectee's injuries is one of the more important and yet most difficult tasks associated with an ejection investigation. A stated primary purpose of the paper is to assist medical offi-cers, pathologists, and other escape system usage investigators to better understand the probable injury-producing mechanism(s) present in each type of system, and hence produce more meaningful and accurate data. The authors argue that careful, detailed investigation (and also general statistical investigation) of "ejection associated" injuries and circumstances often reveals that the assigned causal factors either cannot be applicable or are of extremely doubtful applicability for the specific situations. They say that evidence now being developed suggests strongly that many injuries attributed in the databank as being caused by the escape system in fact occurred before ejection initiation. They argue that aiding and abetting the selection of incorrect causal factors is the "strength-in-numbers" type of legitimacy that many factors have acquired through frequent usage over the years. Miscellaneous, selected statistical data from the paper are noted here: 15 per-cent of ejections result in fatality; thoracic/lumbar vertebral fractures attributed to ejection are seven times as frequent as cervical fractures attributed to ejection; thoracic/lumbar and cervical sprains/strains occur with equal frequencies; there are higher rates of vertebral compression fractures in through-the-canopy ejectees than in jettisoned canopy ejectees.														
Sandstedt, P.	1989	Experiences of rocket seat ejections in the Swedish Air Force: 1967-1987	X	X	X		up to 1200 kph	X		average gross anthro-pometry	X	X	X	X	X	X	X
			SUMMARY / COMMENTS: The paper describes injury experience in 83 nonfatal and nine fatal ejections in the Swedish Air Force from 1967 to 1987. Thirty-nine of the 83 survivors had nontrivial injuries. Thirty-three of the ejection situations were fully within design ejection envelopes, i.e., for "good" flight situations and crew member sitting postures. In those 33 there were six occurrences of fractured spines (18%) while for 22 in which both the flight situation and the sitting posture were good there were also six with spinal fractures (27%). When sitting posture is not considered as a factor, there were 39 ejection situations considered as "good"--i.e., with favorable aircraft attitude, acceleration, and rotation. Of those 39 there were eight with fractured spines (21%). In the other 44, there were nine spinal injuries (20%). Thus, although the author does not make the point, it would seem that it may be deduced from the data that neither crew member sitting posture nor aircraft flight situation is a strong determinant of the likelihood of spinal fracture. The data show clearly, however, a tendency toward fractures higher in the thoraco-lumbar spine for situations with bad aircraft flight situation and/or bad sitting posture. The author excludes from his "fractured spine" statistics all but the 17 he judges to have resulted from ejection forces.														
McCarthy, G. W.	1988	USAF take-off and landing ejections, 1973-85	take-off, landing									X					
			SUMMARY / COMMENTS: This article examines the injury outcome of the 22 take-off and landing USAF aircrew member ejections that occurred from 1973 to 1985. Three of the ejectees were fatalities, i.e., 13.7 percent. Of the 19 survivors there were four, i.e., 21.1 percent, who received major injuries, which in each case were														

Table 8. INJURY PRIORITY ANALYSIS (Subtask 2) Summary of References

REFERENCE			ESCAPE SEQUENCE PHASES OF PERTINENCE				PARAMETERS ADDRESSED				INJURY TYPES ADDRESSED						
Author(s)	Date	Title	Type of Aircraft Maneuver	Crewmember Pre-Escape Positioning	Ejection Boost Forces	Helmet Impact w. Canopy	Air-craft Speed	Severity of Maneuvers	Mission Require-Ments	Crewmember Physiology and Anthro-pometry	Cervical Spine Dis-locations/ Fractures	Thoracic, Lum-bar, & Sacral Dislocations and Fractures	Musculature & Ligamen-tous Sprains and Strains	Joint Disloca-tions	External and Internal Con-tusions and Bleeding	Concus-sions	Burns
McCarthy, 1988 (continued)			compression fractures of the thoraco-lumbar vertebrae. These results are similar to the results for 705 USAF ejectees in ejections above 500 ft in the same years. There, the fatality rate was 11.5 percent and the major injury rate (based on survivals) was 22.8 percent, with most injuries being spinal compression injuries. It was found in the study that non-takeoff/landing ejections below 500 ft are by far the most dangerous. There, the fatality rate is 53.7 percent and the major injury rate is 53.1 percent.														
Rowe, K. W.; Brooks, C. J.	1984	Head and neck injuries in Canadian forces ejections	high-performance				115 to 550 kn	X			X	X	X		X	X	X
			SUMMARY / COMMENTS: Injuries and helmet performance were studied for Canadian Forces aircrew ejections during the period 1972 to 1982. This paper attempts to distinguish between injuries received during ejection and ones received post ejection. Injuries are not tabulated in greater detail than "none," "minor," "major," "fatal," "overall," and "head and neck." Nor are cases of multiple injuries identified. Thus, in the paper the number given for "overall injuries" is the same as the number of ejections studied, viz., 77. Nonetheless, some statistics pertinent to injuries during ejection can be extracted from the narrative. Five fatalities all resulted from terrain impact. Apart from the fatalities there were eight other instances of major injury. Six of the eight occurred during ejection, and four of those involved the head and neck. The other two were a back injury and a leg injury (from patella striking the canopy). Twenty-eight of the 47 occurrences of "minor" injury occurred during ejection. No breakdown is given by minor injury type for those 28, but for all 47 there were 19 that involved head and neck injury, and cuts, abrasions, bruises, and sprains occurred in about half.														
Voge, V. M.; Borowsky, M. S.	1983	Naval aviation statistics and reports of post-mishap head and spine injuries			X						X	X	X	X	X	X	X
			SUMMARY / COMMENTS: This paper analyzes the data base of injuries in all naval aircraft mishaps from 1973 to 1982, including those for prop aircraft and also jet aircraft mishaps in which aircrew did not eject. The results for ejections will be summarized here. The authors give special attention to head and neck injuries, but some data specific to injuries of the thoracic and lumbar spines are given as well. The most common head and neck injuries that occurred during the ejection sequence were contusions, abrasions, and sprains--each occurring in about 25 percent of the injury cases. Fractures/dislocations were the most common head and neck injury diagnoses in fatal ejections, occurring in 49 percent of the cases. Data are tabulated for the distribution of spinal fractures among the cervical, thoracic, lumbar, and sacrum/coccyx regions in ejection incidents. It may be that most of the fractures can be attributed to ejection forces, but the authors do not attempt to distinguish between fractures occurring during ejection and ones occurring post ejection. In ejection incidents that were fatal and in which vertebral fracture(s) occurred, 77% had cervical fractures and 23% had fractures of the thoracic spine and below (without cervical fracture). In nonfatal ejection incidents in which vertebral fracture(s) occurred, 13% had cervical fractures, 55% had thoracic fractures, and 26% had lumbar fractures.														
Desjardins, S. P.; Coltman, J. W.; Laananen, D. H.	1982	Development of improved criteria for energy-absorbing aircraft seats			X					X							
			SUMMARY / COMMENTS: The reported study develops guidance in the design of aircraft seats of improved crashworthiness through use of experiments with both dummies and cadavers. The loading vector is +Z as it is in aircrew member catapult ejection. Since typical peak +Gz magnitudes, pulse durations, and Z-velocity changes are not dissimilar in survivable crashes and ejections, some of the results of the reported study are pertinent to Task 2 of the current study. Results presented could be useful in anticipating and understanding the effects of changing system constants in ejection tests, such as dummy weight, for example. Specifically, the paper describes the effects of changing the following conditions independently of other conditions: a) input pulse shape; b) magnitude of input acceleration; c) velocity change; d) rate of onset of acceleration; e) dummy type; f) dummy percentile; g) cadavers vs. anthropomorphic dummies; h) seat energy absorber limit load; i) load-deflection characteristics of energy absorbers; j) movable seat weight; k) seat frame spring rate; l) seat cushion stiffness; and m) seat orientation.														
Hearon, B. F.; Brinkley, J. W.; Luciani, R. J.; von Gierke, H. E.	1981	F/FB-111 ejection experience (1967-1980). Part 1: evaluation and recommendations			X					effect on shoulder harness angle	vertebral fractures (region of spine not specified)						
			SUMMARY / COMMENTS: F/FB-111 accident ejection data were examined for the period 19 October 1967 to 26 March 1980. There were 100 ejectees (50 ejections) and 80 survivors (80%). Twelve percent of the survivors received vertebral fracture injuries attributed to either restraint retraction or ejection forces. (It was not possible to distinguish between the two.) Fourteen percent received vertebral fractures attributed to ground landing. Vertebral level, or region, was not specified for the injuries. Injuries of other types are not described or numbered in the report.														
Naval Safety Center	1981	Aircrew Automated Escape Systems (AAES) data analysis program symposium, Volumes I, II, III, and IV.	X	X	X	X	X	X	X		X	X	X	X	X	X	X
			SUMMARY / COMMENTS: This four-volume document from a three-day symposium held at the Naval Air Station, Norfolk, Virginia, in October 1981, is 1234 pages in total length. The symposium was sponsored by the Naval Safety Center, and presentations were made by the Naval Air Systems Command (Aircrew Systems Division), Naval Weapons Engineering Support Activity (Systems Analysis), and the Naval Safety Center (Aviation Directorate). The research presented was conducted for the purpose of evaluating or monitoring usage of Automated Airborne Escape Systems (AAES) and AAES performance and maintenance trends. Source data were derived from date files maintained by the Naval Safety Center. All of the research presented related to aircrew ejection, but much of it was not restricted to the catapult or rocket ejection phase of escape; i.e., much of it related to the phases of escape which follow exit from the aircraft. Therefore, much of the material in these volumes is not of pertinence to ejection-phase injuries, to which the current project is restricted. Material from Volumes I, II, III, and IV that has pertinence to injury priority analysis (for the ejection phase of escape) is summarized below for the four volumes separately.														
Naval Safety Center	1981	Aircrew Automated Escape Systems (AAES) data analysis program symposium, Volume I.	-- not of pertinence to Task 2 --														
			SUMMARY / COMMENTS: Volume I contains only statistical information relative to the general circumstances and the general outcome of naval aircrew member ejections. There is no assessment of when during the escapes the resulting injuries occurred. Further, injury results are described only in general terms (i.e., no injury, minor injury, major injury, fatality, lost). This volume contains a large amount of data on distribution of ejections by air speed and seat type, by seat type and pitch angle, and by seat type and back angle, but no information is given about resulting injuries for any of those distributions. The Injury Priority Analysis task of the current study is limited to ejection-phase injuries so nothing in Volume I is pertinent.														

Table 8. INJURY PRIORITY ANALYSIS (Subtask 2)
Summary of References

REFERENCE			ESCAPE SEQUENCE PHASES OF PERTINENCE				PARAMETERS ADDRESSED				INJURY TYPES ADDRESSED						
Author(s)	Date	Title	Type of Aircraft Maneuver	Crewmember Pre-Escape Positioning	Ejection Boost Forces	Helmet Impact w. Canopy	Air-craft Speed	Severity of Maneuvers	Mission Require-Ments	Crewmember Physiology and Anthro-pometry	Cervical Spine Dis-locations/ Fractures	Thoracic, Lum-bar, & Sacral Dislocations and Fractures	Musculature & Ligamen-tous Sprains and Strains	Joint Disloca-tions	External and Internal Con-tusions and Bleeding	Concus-sions	Burns
Naval Safety Center	1981	Aircrew Automated Escape Systems (AAES) data analysis program symposium, Volume II.		many factors discussed	X	X					X	X	X				
Naval Safety Center	1981	Aircrew Automated Escape Systems (AAES) data analysis program symposium, Volume III.	-- not of pertinence to Task 2 --														
Naval Safety Center	1981	Aircrew Automated Escape Systems (AAES) data analysis program symposium, Volume IV.	X	X	X	X					X	X	X	X	X	X	X

SUMMARY / COMMENTS: Much of the data in Volume II is not pertinent to the current study as it either does not relate specifically to the ejection phase of escape or it contains no information about injuries. There is, for example, ejection distribution data by seat type, seat families, and seat groups (jettisoned canopy and through-the-canopy). There is also information about parachute types and results of over-water ejections. Information of pertinence to the current study includes data comparing vertebral injury rates for through-the-canopy and canopy-jettisoned ejections (Martin-Baker and NAMC II seats, 1 Sept 1958 through 31 Dec 1961). Injury rates were about nine times as large for the through-the-canopy ejections, viz., about 35 percent to about 4 percent. (Paravertebral muscle strain/sprain incidence was included with vertebral fractures in the definition used for "injury.") For the through-the-canopy ejection cases of vertebral injury a loose restraint harness was assigned as the cause of about 9 percent of the injuries for the Martin-Baker seats, and 34 percent of the injuries were attributed to "poor position." (The remaining 57 percent were, presumably, thought to result from out-of-envelope aircraft attitude or ground impact.) For all Martin-Baker seat, survived ejections (1958 to 1963), 73 percent of the vertebral fractures were from T8 to T12 and only three percent (two of 63) were in the neck. The percentages are very little different if only through-the-canopy ejections are considered. Other data presented (1969 to 1979), however, show a distribution for through-the-canopy ejections that is shifted away from the lower thoracic region (and the cervical spine, as well) to the midthoracic region. ESCAPAC series ejection seats have relatively greater rates of vertebral fracture in the midthorac and cervical regions. Equipping ESCAPAC seats with powered inertia reels (1969 to 1974) was found to reduce the rate of vertebral fractures by a factor of two while reducing the rate of neck sprain/strain by a factor of six. It is stated that for current (1981) ejection systems, except possibly for through-the-canopy ejections, most ejection-associated vertebral compression fractures are the result of poor vertebral alignment. Causes suggested for misalignment (discussed in detail) include: personal equipment influences, nonstable ejection platform, inadequate thigh support, poor torso restraint, forward torso rotation induced by rear-angled catapult boost acceleration vector, poor seat back support, and upper torso movement. With regard to through-the-canopy ejection it is noted that the head contacting the canopy may induce vertebral misalignment.

SUMMARY / COMMENTS: Volume III deals exclusively with statistics for injuries attributed to windblast, flail, and tumble following complete egress from the aircraft. The Injury Priority Analysis task of the current study is limited to ejection-phase injuries.

SUMMARY / COMMENTS: Volume IV contains detailed statistics about all U.S. Navy ejections that occurred from Jan 1969 to Dec 1979. Injuries in through-the-canopy ejections and jettisoned-canopy ejections are compared in most of the tabular presentations of injuries sustained by survivors. Injury data are presented by body part and by type. The detail of presentation in this document is greater than found almost anywhere else in the literature. It is not clear, however, how fully pertinent the data are to the Injury Priority Analysis task of the current study, as defined, because no distinction is made in the data tables between injuries that occurred before and after complete egress from the aircraft. Nonetheless, a summary of the most often occurring injuries is given here. In the table below both the actual numbers of injuries that occurred and the rates are given. Rates are based on a total of 179 survivors of through-the-canopy ejections and a total of 941 survivors of jettisoned-canopy ejections. For assessing the difference between through-the-canopy ejections and jettisoned-canopy ejections for a particular type of injury, the rates should be compared. Of greater relevance to the current study, however, is comparison of the actual numbers of injuries of different types. Such comparisons can be made for either type of ejection or for the survivors of both types of ejection. The numbers do not necessarily reflect relative overall importance, however, since both minor and major injuries are included here. Thus, while, for example, there were 358 contusions for the 941 survivors of jettisoned-canopy ejections and there were only 15 C1-C7 neck injuries (mostly vertebral fractures), the overall negative consequence of the neck injuries is probably greater. The tables in Volume IV break down the injury numbers by body part. In the table below, numbers are the total number of injuries of a given type (e.g., simple fracture) for all body parts. Note is made here that the distributions over body parts for most types of injuries are similar for through-the-canopy and jettisoned-canopy ejections.

	Through-the-Canopy (179)		Jettisoned-Canopy (941)		Total (1120)	
NON-SPINAL INJURIES	Number (N)	Rate (%) (100xN/179)	Number (N)	Rate (%) (100xN/179)	Number (N)	Rate (%) (100xN/179)
comminuted fracture	8	4%	10	1%	18	2%
simple fracture	23	13%	78	8%	101	9%
dislocation	6	3%	21	2%	27	2%
strain	48	27%	178	19%	226	20%
sprain	8	4%	70	7%	78	7%
concussion	1	1%	5	1%	6	1%
thermal burn	17	9%	40	4%	57	5%
laceration	68	38%	86	9%	154	14%
contusion	133	74%	358	38%	491	44%
hematoma	11	6%	27	3%	38	3%
abrasion	75	42%	172	18%	247	22%
SPINAL INJURIES (all types, mostly fractures)						
C1-C7	3	2%	15	2%	18	2%
T1-T5	15	8%	10	1%	25	2%
T6-T12	23	13%	33	4%	56	5%
L1-L5	13	7%	23	2%	36	3%
C1-T5	18	10%	25	3%	43	4%
T6-L5	45	25%	56	6%	101	9%
Sacrum/Coccyx (mostly contusions)	5	3%	13	1%	18	2%

Table 8. INJURY PRIORITY ANALYSIS (Subtask 2) Summary of References

REFERENCE			ESCAPE SEQUENCE PHASES OF PERTINENCE				PARAMETERS ADDRESSED				INJURY TYPES ADDRESSED						
Author(s)	Date	Title	Type of Aircraft Maneuver	Crewmember Pre-Escape Positioning	Ejection Boost Forces	Helmet Impact w. Canopy	Air-craft Speed	Severity of Maneuvers	Mission Require-Ments	Crewmember Physiology and Anthro-pometry	Cervical Spine Dis-locations/ Fractures	Thoracic, Lum-bar, & Sacral Dislocations and Fractures	Musculature & Ligamen-tous Sprains and Strains	Joint Disloca-tions	External and Internal Con-tusions and Bleeding	Concus-sions	Burns
Fleming, C.	1979	Ejection problems and injuries: their causes, effects and treatments, and suggestions for preventive measures	mostly in spins or turns	re. use of upper or lower ejection handle	cervical and thoracic spine injuries		mostly 350 to 400 kn; maximum of 550 kn	mostly in spins or turns	combat		C4, C5 (no relative rates given)	T4, T5, T6, T8 (no relative rates given)		(post-ejection)			(from fire, not rocket)
			SUMMARY / COMMENTS: Data are from combat ejections which took place mainly during the Yom Kippur War of 1973 in the Israel Air Force (I.A.F.). The only injuries mentioned that are pertinent to the ejection phase of escape are spinal injuries and hand injuries. No indication is given of the relative occurrences or severity of cervical and thoracic spinal injuries.														
Kazarian, L. E.; Beers, K.; Hernandez, J.	1979	Spinal injuries in the F/FB-111 crew escape system		powered inertial reel								mostly midthoracic					
			SUMMARY / COMMENTS: In the years 1970 to 1975, before redesign of the crew escape module, aircrew ejecting from the F/FB-111 had a larger than expected rate of ejection-associated acute spinal trauma. Injuries discussed were all to the thoracic spine. (As there is no mention in the paper of cervical spine injuries, they were presumably absent or nearly absent.) Statistics regarding the rate and severity of injuries are given. Injuries were primarily of two types: a) retro-hyperflexion injury of the midthoracic spinal column; b) hyperflexion injury of the thoracic spine. The second of these results from ground impact by the escape module. The first occurs as an aspect of the ejection procedure, viz., activation of the powered inertial reel, which forces the crewman into an erect seated position prior to catapult thrust. The occurrence of this injury is a function of seat geometry and harness configuration.														
Norman, R. W.; Bishop, P. J.; Pierrynowski, M. R.; Pezzack, J. C.	1979	Aircrew helmet protection against potential cerebral concussion in low-magnitude impacts				X										X	
			SUMMARY / COMMENTS: In experiments using a Hodgson headform head accelerations were measured for crown impacts with two types of helmets, the Gentex DH-151 and the Gentex 411, as well as without a helmet. The results therefore have pertinence to study of potential for concussive injury during through-the-canopy aircrew ejections. The protective mechanism of the DH-151, a "contact" helmet, is based largely on liner and shell disintegration during impact. The "411" helmet protects through energy absorption by the helmet liner and a suspension. While both helmets provided adequate protection for impact velocities of up to 5 m/s (on the basis of Gadd Severity Index values), neither protected from moderate to severe concussion for impact velocities above 7 m/s. Further, it was found that a (head-vs-canopy) double hit greatly reduces the protection provided by both helmets.														
Kazarian, L. E.	1978	Identification and classification of vertebral fractures following emergency capsule egress from military aircraft		geometry and support of restraint system; powered retractor	X					torso height	relatively uncommon	mostly T11-L2 for open ejection seat; mostly midthoracic for crew escape module					
			SUMMARY / COMMENTS: Data analyzed are for military aircraft crew escape module ejections between 1967 and 1976. Powered restraint system retraction is the primary cause of upper and midthoracic spinal column injuries. The most common ejection injuries there are due to indirect forces in which there occurs hyperextension or hyperflexion of the spinal column within the confines of the support and restraint system. The powered retraction, individual torso height, and restraint system geometry are causative while +Gz ejection forces are only contributive. Ejection forces are important in open seat ejections, in which spinal column injuries are predominantly between T11 and L2.														
Zenobi, T. J.	1978	Development of an inflatable head/neck restraint system for ejection seats		X	X						X		cervical			X	
			SUMMARY / COMMENTS: The author states that data from the Naval Safety Center for over 1300 Navy aircraft ejections during 1967-1974 indicate that neck injuries ranging from minor to critical occurred in approximately eight percent of the ejections. The author does not state the percentages attributed separately to ejection forces and parachute opening forces. No information about other injuries is given.														
Kazarian, L. E.	1977	F/FB-111 escape injury mechanism assessment		powered inertial reel								mostly midthoracic					
			SUMMARY / COMMENTS: This is the full research report on which the 1979 Kazarian, et al., reference is based. The problem of aircrew spinal column injuries in ejections of the F/FB-111 crew escape module is addressed. (See above.) Selected additional information of note: a) the most common region of spinal column injury in open seat ejections is the same as in helicopter crashes--viz., approximately T10 to L2; b) open seat ejection injuries are uncommon in the C5 to T7 region of the spinal column.														
Auffret, R.; Delahaye, R. P.	1975	Spinal injury after ejection; Lesions vertebrales apres ejection	X	X	X	X				no relationship to injury found	X	X					
			SUMMARY / COMMENTS: Researchers representing the military services of the U.S., France, West Germany, Greece, Italy, and the United Kingdom contributed statistical data regarding injuries experienced in aircrew ejections with "modern" (1975) flight seats. The authors stress that spinal fractures are the most frequently encountered injuries, occurring in 10 to 47 percent of surviving pilots depending on the data base examined. Thirty-seven percent of all fractures occurred at T12 or L1. This study found that 64 percent of the fractures were in the thoracic spine, 32 percent were in the lumbar spine, and only 4 percent were in the cervical spine. It is noted that multiple thoracic fractures are common in ejections. The report includes an extensive discussion of the mechanics of occurrence of spinal fractures. All observations are relevant for ejection seats that produce accelerations of less than 20 G for 200 to 500 ms and a rate of onset of less than 200 G/s. It is stated that the most important factors affecting likelihood of injury are the posture and position of the pilot at the moment of ejection. The pilot should be seated erectly and should be restrained by a harness that does not allow excessive freedom of movement of the torso (especially in flexion). The harness should be tight enough to hold the pilot in position even in high-G maneuvers since abnormal flight configurations may well exist at the instant of ejection. The seat pan angle should be such that the angle between the torso and the thigh is 135 degrees for proper alignment of the thoracic vertebrae. A hard seat cushion helps to reduce compressive loads.														

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Author(s)	Date	Title	Type of Aircraft Maneuver	Crewmember Pre-Escape Positioning	Ejection Boost Forces	Helmet Impact w. Canopy	Air-craft Speed	Severity of Maneuvers	Mission Require-Ments	Crewmember Physiology and Anthro-pometry	Cervical Spine Dis-locations/ Fractures	Thoracic, Lum-bar, & Sacral Dislocations and Fractures	Musculature & Ligamen-tous Sprains and Strains	Joint Disloca-tions	External and Internal Con-tusions and Bleeding	Concus-sions	Burns
Delahaye, R. P.; Carre, R.; Auffret, R.	1975	Fractures of the spine in helicopter accidents (examination of 25 cases)			pertinent							mostly T10 to L2					
			SUMMARY / COMMENTS: Various causes of helicopter crashes are discussed. During the period 1961 to 1966 in the French Air Force 8.8 percent of pilots in helicopter crashes suffered fractures of the spine. It is not stated whether these are surviving pilots or all pilots involved in helicopter crashes. It may be noted, however, that the spinal fracture rate is less than the 10 to 47 percent cited by the same authors for (survived) pilot ejections in another 1975 paper (above). The most likely cause of the difference is possibly simply that the +Gz accelerations, on average, in (survivable) helicopter crashes are less than in pilot ejections. The authors note that--despite different seats and restraint systems in helicopters and jet aircraft--the distributions of fractures by position along the spine are similar. Data reported at the XVIth Congress of Aerospace Medicine are cited, which indicate T10 to L2 as the most frequent position for spinal fracture in both ejections and helicopter crashes. This similarity is also noted by Kazarian (1975, 1977, 1979).														
Kazarian, L. E.	1975	Standardization and interpretation of spinal injury criteria and human impact acceleration tolerance			X												
			SUMMARY / COMMENTS: This article primarily describes in detail the types and severities of vertebral body trauma of all sorts without regard to cause. It does include discussion, however, of the distribution of spinal column fractures that result from ejections from aircraft. The author notes that the most common site is the T12 to L1 region of the spine. He further notes that the upper/mid thoracic region is the least common site for spinal fractures resulting from aircraft or helicopter crashes and ejections, but the data he presents show that, contrarily, the cervical region of the spine has the least incidence of fractures from those causes.														
Rotondo, G.	1975	Spinal injury after ejection in jet pilots: mechanism, diagnosis, followup, and prevention		X	X	X				effects of lumbago, discal prolapse, arthrosis, ischialgia, kyphotic & scoliotic deviations, spondylolysis, spondylolisthesis	none observed	observed at T7-L4					
			SUMMARY / COMMENTS: The occurrence of spinal injury after ejection was studied for 100 ejections by military and civil Italian jet pilots in the time frame 1955-1975. Forty-seven pilots were uninjured. Eleven pilots died, but all deaths were judged to have resulted from ground impact; the presence of spinal injuries in the fatalities was not considered in the study. Of the remaining 42 injured pilots, 15 had spinal fractures and 27 had some other type of trauma. Only the cases with spinal injuries are discussed by the author. The author believes that nearly all of the spinal injuries among the 15 were caused by ejection forces. No cervical spine fractures were found (although there may have been cervical soft tissue damage in some of the 27 non-fracture cases). There were 23 thoracic and lumbar vertebral fractures among the 15 cases with fractures. The distribution of fractures was nearly uniform over the entire range of occurrence, from T7 to L4, except for T12 and L1, where the rate of occurrence was nearly four times as great.														
Brinkley, J. W.; Shaffer, J. T.	1971	Dynamic simulation techniques for the design of escape systems: current applications and future Air Force requirements		re. use of ejection handle and belt tightness	X		w.r.t. altitude				not separately identified	X					
			SUMMARY / COMMENTS: This paper largely discusses use of a one degree-of-freedom dynamic response model for predicting the probability of spinal injury in aircrew ejections. The authors comment, however, on observational data with regard to injury rates and the effects of various factors that bear on aircrew exposure to a dynamic environment. Note is made that ejection boost acceleration is the primary cause of major injuries related to ejections. (The second largest cause is stated to be post-ejection limb flailing, accounting for 12 percent of the total number of major injuries. Only five major injuries resulted from parachute opening shock in a study of 384 ejections.) A study of F-4 ejections attributed 84 percent of all spinal injuries to ejection forces. During the period 1966 to 1968 the spinal injury rate for F-4 ejections was 41 percent while it was approximately four percent for other U.S. Air Force ejection seats. The large difference was attributed partly to the fact that the F-4 catapult acceleration vector is not aligned with the crew member's vertebral column. Additionally, a lap belt that was difficult to tighten and a suboptimal seat contour were identified as problems; correction of those two design elements together with a reduced charge catapult lessened the F-4 spinal injury rate from 41 percent to eight percent. The importance of body support and restraint systems was suggested in the paper but not discussed in detail.														
Nuttall, J. B.	1971	Emergency escape from aircraft and spacecraft		erect (need high handle)	X	X	X				X	X					
			SUMMARY / COMMENTS: This article is a thorough discussion of all aspects of emergency escape from aircraft and spacecraft. One section covers the subject of escape by ballistic catapult or rocket seat ejection in considerable depth, although with scant data for injury statistics. No types of injury other than spinal fractures are mentioned as resulting from ejection seat accelerations. The region T11-L2 is identified as the most vulnerable part of the spine in terms of number of +Gz-related fractures. The author cites a study of pre-1957 nonfatal ejections in which it was estimated that only 2.2% resulted in spinal fractures caused by seat ejection forces. (The cited report was "The Problem of Back Fractures During Ejection from USAF Aircraft," 1957.) In another cited study (Chubb, et al., 1965) only 3.8% of ejections caused spinal fractures. In that study the spinal injury rate for crew members known to be improperly positioned during ejection was 13 times as large as the rate for erectly positioned crewmembers. Both studies cited by the author indicate that the most important factor in preventing spinal fracture is an erect posture with the head and buttocks pressed firmly back into the seat. The Chubb study indicates also that it is important for the thrust vector of the seat to be parallel to the spinal column or forward from it to prevent anterior-lip compression fractures in the cervical and upper thoracic spine.														
Orne, D.; Liu, Y. K.	1971	A mathematical model of spinal response to impact			X												
			SUMMARY / COMMENTS: This paper presents a discrete-parameter mathematical model of the human spine and simulations of pilot spinal column response in ejections at														

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Orne and Liu, 1971 (continued)			10 to 20 peak +Gz. The authors make note of results compiled by Moffatt and Howard of vertebral fractures in over 1000 U.S. Navy and Air Force aircraft ejections from 1959 to 1967. No data for cervical spine injuries are given. The preponderance of thoracic spine fractures occur from T7 to T12, with a distinct lessening from T6 to T1. Fractures from L1 to L4 were uncommon.														
Shannon, R. H.	1971	Operational aspects of forces on man during ejection/extraction escape in the US Air Force 1 Jan 1968 - 31 Dec 1970		X	X					age and weight		most spinal fractures were from T11 to L2					
			SUMMARY / COMMENTS: U.S. Air Force ejections from 1 January 1968 to 31 December 1970 were studied. In 52 of the total of 468 ejections (i.e., 11 percent) the crewmember received major (49) or fatal injuries (3). In 62 percent of major-injury cases ejection forces were judged responsible for the primary injuries. Windblast and parachute opening shock were identified in only 28 percent and 10 percent, respectively, of the cases. All but two of the major injuries associated with ejection force were compression fractures of the vertebral column. The majority of the fractures were in the T11 to L2 region. The (major) injury rate for straight ballistic catapult systems was 12 percent for all nonfatal ejectees. The rate was eight percent for rocket-assisted systems. Body position at the time of ejection was found to be the factor that correlated most strongly with injury rate. The injury rate was four percent among optimally seated ejectees (head and buttocks back into the seat) and 31 percent among improperly positioned ejectees. Proper pre-ejection position was found to be of even greater importance among older (30+) crewmembers. No significant differences were found in injury rates for crewmembers of different weights when other factors, such as pre-positioning, were accounted for.														
Symeonides, P. P.	1971	Some observations on compression fractures of the spine in ejected Greek pilots		excessive tightening of belts causes flexion	X							X					
			SUMMARY / COMMENTS: Injuries resulting from Greek pilot ejections in the ten years 1960 to 1969 were studied. Of 33 surviving ejectees six (18 percent) sustained compression fractures of the spine. Such injuries were the most important type sustained by pilots in successful ejections. (The number, assigned cause, and injury type of fatalities are not mentioned in the paper.) All vertebral fractures (12 among the six injured pilots) were from T10 to L2. The author states a finding that injury rates were found to be higher among pilots who tightened their shoulder-buttock belts excessively so that the shoulders were forced down, causing in fact greater flexion and vulnerability of the spine.														
Henzel, J. H. 53	1967	The human spinal column and upward ejection accelera-tion: an appraisal of biodynamic implications		X	X					X		X					
			SUMMARY / COMMENTS: This report contains a thorough summary of early work (1939-1950) on ejection systems by Germany, England, Sweden, and the United States. It also discusses work done in the United States through 1967. The report deals primarily with the spinal column vertebral body and intervertebral disc response during ejection. The author notes that other spinal injuries also occur--dislocation, posterior arch disruption, and cord transection--but that they are not discussed in the report. Thoracic and lumbar vertebral injury data are given from two cited ejection seat studies. Both are for aircrew surviving ejection with the Martin-Baker seat (18-21 G peak). British data (Fryer, 1961) from 1949 to 1960 show that most of the injuries occurred from T10 to L1 and that injuries above T5 and below L2 were uncommon. British, Swedish, and U.S. data (Jones, et al., 1964) show very similar results. The most common injury site was T12. The most common lumbar injury site was L1. Approximately 80% of thoraco-lumbar vertebral fractures occurred in the thoracic region. The author states that a great portion of ejection-incurred spinal injuries result from abnormal ejectee posture, unsuspected congenital spinal weakness, and dynamic spinal overloading occurring secondary to either "overshoot" or improperly arranged restraint and stabilization systems.														
Higgins, L. S.; Enfield, S. A.; Marshall, R. J.	1965	Studies on vertebral injuries sustained during aircrew ejection	X	X	X	X	X	X	X	X	(X)	X					
			SUMMARY / COMMENTS: Literature available in 1965 on ejection-related vertebral injuries in aviators was surveyed. Parameters associated with the pilot, aircraft, and ejection-seat system are evaluated in the light of their trends and relative significance in contributing to ejection-caused vertebral injury. Whether or not the authors made an a priori assumption that vertebral fracture injuries are the injuries of greatest importance in ejections, the study does not in fact consider any other type of injury. The primary data cited that pertain to the incidence of vertebral injury in ejections do not include information on cervical fractures. These are the data of Jones, et al. (1964), for U.S., British, and Swedish ejections. By far the most common injury site is the lower thoracic spine. (See Jones, et al., 1964.) Regarding the significance of various parameters to vertebral injury in aircraft ejections, the authors reach the following conclusions. Pilot age: greater likelihood of injury for age greater than 24. Pilot height/weight: no evidence for relationship to likelihood of injury. Pilot training: proper body position and execution of ejection procedure reduce injuries. Aircraft speed: 25% fatality rate below 500 kn, 65% major injury or fatality above 500 kn. Aircraft altitude: does not affect the vertebral injury rate. Aircraft attitude: important if the pilot is improperly seated because of acceleration forces and/or a loose harness. Ejection method: pre-1965, the vertebral injury rate for ejections through the canopy was 8.5 times the rate for ejections with a jettisoned canopy. Ejection seat: about the same rates of vertebral injury incidence for various seats in use. Ejection velocity: significant increase in injury incidence for 80 ft/s ejections compared with 60 ft/s. Ejection seat pack: lower rates of onset of acceleration (desirable) can be effected by the elastic properties selected for pack or cushion. Seat restraint harness: important for maintaining normal vertebral column alignment during ejection. Ejection axis: should be parallel to spinal axis.														
Jones, W. L.; Madden, W. F.; Luedeman, G.	1964	Ejection seat accelerations and injuries										X					
			SUMMARY / COMMENTS: Thoracic and lumbar vertebral injury data are described for aircrew surviving ejection with the 18-21 G Martin-Baker seat during 1958 to 1963. The data from British, Swedish, and U.S. data bases show that most of the injuries occurred from T10 to L1 and that injuries above T5 and below L2 were uncommon. The most common injury site was T12. The most common lumbar injury site was L1. Approximately 80% of thoraco-lumbar vertebral fractures occurred in the thoracic region.														

Table 9. TRAUMA ASSESSMENT CRITERIA (Subtask 3)
Summary of References

REFERENCE			INJURY CRITERIA					BIOMECHANICAL PROPERTIES				SUMMARY / COMMENTS
Author(s)	Date	Title	Cervical Spine Dislocations and Fractures	Thoracic, Lumbar, and Sacral Spine Dislocations & Fx	Musculature and Ligaments	Head	Other	Cervical Spine	Thoracic, Lumbar, and Sacral Spine	Head	Other	
Hamalainen, O.	1993	Flight helmet weight, +Gz forces, and neck muscle strain			pertinent							The study reported investigated the effect of flight helmet weight on cervical erector spinae muscle strain under +Gz loadings. Helmet masses were about 1.9 kg and 1.3 kg. EMG responses, normalized by MVC's (maximal voluntary contraction), were measured for two pilots during 16 flights consisting of a series of maneuvers that produced sustained loadings of up to +7.0 Gz. The results show that a lower weight flight helmet reduces muscle strain during the most stressing flight maneuvers, at least in some pilots. The highest difference between the helmets during stabilized bank under +4.0 Gz and +7.0 Gz was less than 2.0% and 4.0% of MVC, respectively. It was concluded that reducing the weight of the flight helmet is only of limited value in preventing inflight acute neck pain and related problems. Fatigue but no muscle, or other, pain was reported by the pilots in any of the flights.
Buhrman, J. R.	1991	Vertical impact tests of humans and anthropomorphic manikins						pertinent	pertinent			Small and large prototype Advanced Dynamic Antropomorphic Manikins (ADAMs) and CG-5 and CG-95 GARD manikins were tested in +Gz up to 24 g, and human subjects were tested at up to 10 G. No mechanical or biomechanical property data or injury criteria are given in the report, but findings have pertinence to design of any new manikin that is to be used in assessing likelihood of injuries in ejections. The large ADAM and the small and large GARD manikins demonstrated inconsistent simulation of the gross dynamics of human response. Only the small ADAM yielded good results for gross dynamics, i.e., accelerations, velocities, and displacements. The implication is that only the small ADAM would be capable, in theory, of predicting with reasonable accuracy the body linkage forces and moments that might be used to predict injury. Thus, it seems of importance to establish what differences in mechanical properties or other design parameters for the four dummies are responsible for their respective good or poor performances. The study did not determine specific reasons for the poor performance of the large (97th-percentile) ADAM, but the report states that it is not suitable for simulating human dynamic response.
Kallieris, D.; Mattern, R.; Miltner, E.; Schmidt, G.; Stein, K.	1991	Considerations for a neck injury criterion	X	T1 to T4	X							Twenty-three frontal car crashes were conducted with a vehicle crash barrier and also fourteen car-to-car lateral collisions. Cadaver subjects were used. High-speed film was analyzed to determine head and neck angular and translational displacements, velocities, and accelerations. Neck injuries resulted primarily from forward and lateral flexion of the head/neck. The highest correlation of any determined response with severity of neck injury was maximum head translational acceleration in the direction of the trajectory (i.e., along the path). Above a value of 21 G for this response, there was injury in every case. Compression fracture of the cervical vertebrae was uncommon. Rupture of the intervertebral disc was the most common of all types of cervical injuries observed.
Nightingale, R. W.; Myers, B. S.; McElhaney, J. H.; Doherty, B. J.; Richardson, W. J.	1991	The influence of end condition on human cervical spine injury mechanisms	for combined flexion and axial loading					axial stiffness				The passive combined flexion and axial loading responses of the unembalmed human cervical spine were measured. Different end conditions (unconstrained, rotational constraint, and full constraint) greatly influenced the risk of injury, the failure mode, and the observed axial load to failure. The results suggest that safety equipment and injury environments should be designed to minimize the degree of imposed constraint on the head. In particular, systems which tend to "pocket" may produce an enhanced injury potential. The axial load to failure for lower cervical bilateral dislocation is significantly lower than the axial load to failure for vertebral compression type fractures.
Society of Automotive Engineers, Inc.	1991	SAE Information Report, SAE J1460 MAR85: Human Mechanical Response	lower bound and marginal-injury criteria for flexion and extension					flexion and extension response data				This is a report prepared by the Human Injury Criteria Task Force of the Human Biomechanics and Simulation Subcommittee of SAE. Its stated purpose is to provide a first-generation version of a standardized SAE document to define human mechanical response characteristics. The primary biomechanical system properties pertinent to experimental estimation of injury potential in aircrew ejections are properties of the cervical and thoraco-lumbar spinal columns. No data are given in this report for properties of the thoraco-lumbar spine. No axial or shear data are given for the neck, but torque-deflection data from Mertz and Patrick (1972) are given for full-neck flexion and extension. The energy restitution coefficient for both flexion and extension is about 0.5.

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Society of Automotive Engineers, Inc.	1991	SAE Information Report, SAE J885 JUL86: Human Tolerance to Impact Conditions as Related to Motor Vehicle Design	failure loads									This is a report prepared by the Human Injury Criteria Task Force of the Human Biomechanics and Simulation Subcommittee of SAE. Its stated purpose is to assist the automotive safety designer and tester by providing quantitative data on the strength of the human body under impact loading conditions. No data pertinent to the types of injuries that occur in aircrew ejections are presented in the report except for some for the neck. Most neck data given are for subinjury response of volunteer subjects (Mertz and Patrick, 1972; Patrick and Chou, 1976). Injury level data for axial neck compression (Culver, et al., 1978) from direct S-I head impacts of cadavers are given. The authors comment that since the measured forces were dynamic head loads (for a padded impactor), the corresponding axial compressive neck loads could be smaller due to head mass inertial effects. In the cited cadaver tests peak loads of less than 1560 lb usually did not produce neck fractures. In eleven tests the lowest load which produced vertebral process fractures was 1060 lb. That test also produced some lateral lip crush of the C5 body. The lowest load which produced significant crush of any vertebral body (C5) was 1620 lb. A test with an S-I head load of 1990 lb crushed the C3-4, C4-5, and C5-6 discs, fractured two transverse processes, and severely crushed the T2 vertebral body. Note is made that fractures can occur at on the order of half these loads if the head, neck, and torso are not axially aligned. With regard to shear force injuries in the neck, the authors note that the upper part of the neck (occipital condyles to C2) is most subject to injury. An implication is that it is important to measure shear force in the upper neck in manikin tests. This SAE report does not, however, give injury criteria data for the neck in shear.
Association for the Advancement of Automotive Medicine	1990	The abbreviated injury scale - 1990 revision	pertinent	pertinent	pertinent	pertinent						This document (and earlier versions) was developed by AAAM (and previously the American Medical Association, AMA) to provide researchers with a numerical method for ranking and comparing injuries by severity and to standardize the terminology used to describe injuries. The injury severity code defined and described is the Abbreviated Injury Scale, i.e., AIS. This scale has been applied mostly in automotive medicine. All types of internal and external, impact-related injuries for all regions of the human body are dealt with. No injury criteria are given. Rather, injury severities are described on the following scale: 0=none, 1=minor, 2=moderate, 3=serious, 4=severe, 5=critical, 6=maximum.
Chandler, R. F.	1990	Occupant crash protection in military air transport		pertinent								This report is exhaustive in discussion of research conducted from 1940 to 1990 in development of aircraft seats and restraint systems with improved capability to protect occupants of aircraft involved in a crash. The report does not discuss aircrew ejection systems. Nonetheless, since +Gz accelerations are an important consideration in aircraft crashes, research on seats designed to be maximally protective in crashes should have some pertinence to design of ejection seats. If redesign of ejection seats is suggested by work with a test manikin of improved capability to assess likelihood of +Gz related injuries, this report should be reviewed.
Freivalds, A.; McCauley, D.S.	1990	Biodynamic simulations of helmet mass and center-of-gravity effects	neck torques and head/neck angle			HIC and Gmax						Ejections were simulated with a computer model with different helmet weights and centers of gravity and also different initial resting angles of the head and neck. Regressions were determined for the parametric dependence of head and neck dynamic responses, including indicators of injury potential, on C.G. offset, initial head rest angle, and total head-helmet mass.
Lankarani, H.; Ma, D.; Ermer, G.	1990	Biodynamic simulations of an aircraft pilot/passenger in various crash environments					knee (no injury data)					A simple multiple-segment model of the human body was used to examine dynamic response to +Gz ejection seat acceleration. The constraint forces transmitted to the pelvis and to the knees were calculated among other responses. The largest peaks corresponded to a secondary vertical impact of the pilot on the seat. The simulation results are of questionable value in the study of ejection injuries because a single mass is used to represent the head and torso. Values for mechanical system parameters are not given.
Brinkley, J. W.; Specker, L. J.; Mosher, S. E.	1989	Development of acceleration exposure limits for advanced escape systems		fracture probability			3-axis					The paper describes a method for predicting the probability of injury due to combined accelerations in X, Y, and Z axes. Determined probabilities are based on limit values for independent plus and minus X, Y, and Z accelerations derived from human impact data bases. Aircrew member response in X,

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Brinkley, et al., 1989 (continued)												Y, and Z is estimated from a simple three-axis model, with inputs being from ejection seat accelerations. This technique is called the acceleration exposure limit method. The paper also summarizes the results of impact tests accomplished with volunteers. The acceleration exposure limit method is an extension of the previously used, +Z axis, Dynamic Response Index (DRI) model. The accelerations are presumed to have their greatest deleterious effect when acting at a specific "critical point." For the described study that point was defined as the center of mass of the upper torso. Injury probabilities are estimated from the computed accelerations of that point. A mass at the critical point is attached independently by three linear, spring-damper systems to the ejection seat, which is assigned three linear acceleration components and an angular velocity. Nondimensional dynamic responses of the critical point mass are calculated by dividing the X, Y, and Z accelerations by the previously described acceleration limit values. A time-varying injury-risk measure is calculated as the square root of the sum of the squares of the three nondimensional accelerations. Separate values of this measure are calculated for low-, moderate-, and high-risk limit accelerations. The escape system occupant is considered to have exceeded a specified injury-risk level if this "injury-risk criterion" has a magnitude greater than one. The limit acceleration G's used in the study (and, for -Z, determined by the study) were: Low (+X=35, -X=-28) (Y,w/o side panels= 14; Y,w.side panels= 15) (+Z=15.2, -Z=-13.4) Mod (+X=40, -X=-35) (Y,w/o side panels= 17; Y,w.side panels= 20) (+Z=18.0, -Z=-16.5) High(+X=46, -X=-46) (Y,w/o side panels= 22; Y,w.side panels= 30) (+Z=22.8, -Z=-20.4)
Frisch, G. D.; Kinker, L. E.; Frisch, P. H. 57	1989	Measurement techniques, evaluation criteria and injury probability assessment methodologies developed for Navy ejection and crashworthy seat evaluations	pertinent	pertinent		pertinent						The first generation of biofidelic manikins, designated as BFM1, is described. The primary use of the manikin is intended to be for evaluation of Navy ejection and crashworthy seat systems. The manikin is therefore instrumented with sensors that allow the determination of the dynamic response of the respective segments of interest at locations corresponding to high incidence of injury in ejections and survivable crashes. Thirty-six channels of dynamic response data are collected in tests. The manikin has three linear and three angular acceleration sensors in the head, in the upper thorax, and in the pelvis; additionally, six-axis load cells at the head/neck, neck/thorax, and lumbar spine/pelvis junctions measure a total of nine force and nine moment components. Thus, in addition to inertial response of the manikin, the system determines flexion, extension, and lateral bending moments, and shear and axial forces. The authors allude to data analyses that allow identification of injury mechanisms and probabilities. Intended modifications to the neck are described for a second generation manikin, the BFM2. In describing tests conducted at the Supersonic Naval Ordnance Research Track the authors note the possible production of injury from the post-ejection axial tension response of the neck which can result from large -Gx seat accelerations in transonic ejections. Injury criteria data for use in interpretation of experimental response data are not described except that it is noted that 550 lb of axial tension in the neck approaches the injury level.
McElhaney, J. H.; Doherty, B. J.; Paver, J. G.; Myers, B. S.; Grey, L.	1989	Flexion, extension and lateral bending responses of the cervical spine	failure levels for bending moments and axial loads		ligament failure with failure of cervical bodies			bending stiffnesses in various modes of loading				The 1988 paper by the same authors, "Combined Bending and Axial Loading Responses of the Human Cervical Spine," is very nearly the same as this one. See the summary of the 1988 paper, below.
Oosterveld, W. J.; Kortschot, H. W.; Kingma, G. G.; de Jong, H. A. A.; Saatci, M. R.	1989	Electronystagmographic findings following cervical injuries			cervical whiplash syndrome	headache, vertigo, etc.						This paper details 13 types of complaints of 173 patients with cervical whiplash syndrome resulting from auto accidents--mostly from hyperflexion of the neck. All findings should, however, have pertinence as well to injuries of like type suffered by aircrew during severe maneuvers or emergency egress. Musculature tension in the neck can contribute nearly as much load on the neck as the weight of the head. Neck muscles require 5-20 ms for activation, which may be a longer time than the duration of a dynamic event causing head/neck flexion. During cervical whiplash flexion in a noncritical auto accident, the medulla oblongata and the brain stem can be stretched by as much as 5 cm. The 13 elements of cervical whiplash syndrome identified and discussed in the paper are as follows, with their rates of occurrence in parentheses: unconsciousness (30%), headache (88%), cervico-brachialgia (pain) (94%), vertigo (79%),

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Oosterveld, et al., 1989 (continued)												tiredness (68%), memory difficulties (31%), difficulty in concentrating (28%), depression (22%), irritability (9%), tinnitus (36%), visual disturbances (24%), hearing disturbances (12%), and decreased (sic?) alcohol intolerance (16%).
Pintar, F.A.; Yoganandan, N.; Sances, A., Jr; Reinartz, J.; Harris, G.; Larson, S. J.	1989	Kinematic and anatomical analysis of the human cervical spinal column under axial loading	axial failure loads and deflections					axial loading curves (compression)				Seven fresh human cadaveric head-neck complexes were prepared, and six-axis load cells were placed at the proximal and distal ends of the specimens to document the gross biomechanical response. The preparations were loaded axially to failure at a rate of 2 mm/s. At failure the preparations were deep frozen in the compressed state to preserve tissue alterations. Failure loads ranged from 1355 N to 3612 N for the seven preparations while deflections at failure ranged from 9 to 37 mm. Strains at failure ranged from approximately 0.04 to 0.26. Axial compression loading curves were linear with a slope of 1700 N/cm (average) through about 10 mm of deflection (for each specimen). Upper cervical injuries were observed under compression-extension modes while lower cervical injuries occurred under compression-flexion modes.
Privitzer, E.; Kaleps, I.	1989	Effects of head mounted devices on head-neck dynamic response to +Gz accelerations	NIP criterion	SIF criterion								Computer simulations of ejectee response to +Gz accelerations were conducted for the primary purpose of studying the inertial loading effects of Head Mounted Devices (HMD) on aircrew head-neck-spine dynamic response. The computer model used was HSM, a highly discretized, 3-D representation of the human head, neck, and torso structure. The model uses a Spinal Injury Function, SIF, which makes use of experimental compressive failure data of human thoraco-lumbar vertebrae, to predict the probability of injury, by level, along the thoraco-lumbar spine. A Neck Injury Parameter, NIP, is defined similarly. SIF and NIP values of 1.0 at any vertebral level correspond to a 50 percent likelihood of vertebral body compressive failure due to combined axial compression and bending at that level. The authors state that the injury prediction capability of the model has been validated using operational ejection data.
Schall, D. G.	1989	Non-ejection cervical spine injuries due to +Gz in high performance aircraft	pertinent									Eight non-ejection cervical spine injuries of aircrew members of F-15 or F-16 aircrew are documented. All are attributed to +Gz forces during high performance maneuvers. They include two compression fractures, three left herniated nucleus pulposus, one fracture of a spinous process, one interspinous ligament tear, and one myofascial syndrome. These results are significant with regard to trauma assessment in ejection cases in that they make clear the importance of conservative definition of neck injury criteria with respect to critical loads and moments. In general it may be expected that neck loadings during ejections necessary during defensive or offensive maneuvers, or otherwise high G-loading situations, will be greater than loadings under which the cervical spine injuries reported in this study occurred. The author notes that cervical fractures can occur during flexion and extension at approximately half the axial load required to cause fracture in the absence of flexion or extension.
Yoganandan, N.; Haffner, M.; Maiman, D. J.; Nichols, H.; Pintar, F. A.; Jentzen, J.; Weinshel, S. S.; Larson, S. J.; Sances, A., Jr.	1989	Epidemiology and injury biomechanics of motor vehicle related trauma to the human spine	pertinent									This paper includes no injury criteria data or biomechanical properties data pertinent to design of a manikin to be used in ejection system testing. It does, however, contain auto accident-related spinal injury information that should be considered in the use of manikin test data and in the design of escape systems. The paper examines a large amount of crash victim clinical data and accident data base information. The distribution of spinal injury types is different from that for aircrew member ejections, but it is very clear from the findings that emphasis should be placed on reduction of flexion-compression loadings of the cervical spine and shear loadings at the craniocervical junction. The former is responsible for nearly all cases of complete and incomplete quadriplegia and the latter is responsible for nearly all spinal-column related deaths.
Yoganandan, N.; Sances, A., Jr.; Pintar, F.	1989	Biomechanical evaluation of the axial compressive responses of the human cadaveric and manikin necks	in axial compression					axial stiffness				The neck axial compressive response of human cadaveric preparations was determined under various loading rates and compared with responses from similar tests with a 50th percentile Hybrid III manikin. Cadaveric tests were conducted with intact cadavers, with head-cervical spine specimans stripped of muscle and fat, and with ligamentous cervical column specimans (C2 to T2). Cervical spine fractures occurred in the three tests with intact cadavers at 1512 N, 2936 N, and 1868 N. Axial compressive stiffnesses in the experiments with the head and neck and the neck alone

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Yoganandan, et al., 1989 (continued)												were found to vary from 1.49 to 3.94 kN/cm for different cadaveric preparations. (All of those experiments were for loading rates of 2.54 mm/s.) The authors note that they and other researchers find stiffnesses to be dependent on loading rate but with only moderate sensitivity. For example, for an isolated human cadaver spine, loading rates of 1.27 to 640 mm/s give stiffnesses of 1.18 to 1.93 kN/cm.
McElhaney, J. H.; Doherty, B. J.; Paver, J. G.; Myers, B. S.; Gray, L.	1988	Combined bending and axial loading responses of the human cervical spine	failure levels for bending moments and axial loads		ligament failure with failure of cervical bodies			bending stiffnesses in various modes of loading				The lateral, anterior, and posterior passive bending responses of the human cervical spine were investigated using unembalmed cervical spinal elements obtained from cadavers. Bending stiffness was measured in six modes ranging from tension-extension through compression-flexion. Loads and moments at failure were also determined. Bending stiffnesses were found to be on the order of one-thirtieth the corresponding stiffnesses for the Hybrid III dummy neck. This is said in the paper to be expected because the performance requirements of the Hybrid III were based on human volunteer data, which include the effect of tensed neck musculature. End conditions were found to have a large effect on measured bending stiffness, with values being eight times as large for fixed-pinned conditions as for pinned-pinned conditions. Axial load is found to be a poor indicator of the type and magnitude of failure stresses.
Vanderbeek, R. D.	1988	Period prevalence of acute neck injury in U.S. Air Force pilots exposed to high G forces			spasms, strains, torticollis		sensory & motor deficits					Over 50 percent of 437 pilots of high performance aircraft surveyed by means of an anonymous questionnaire stated they had some type of (unreported) acute neck injury in the preceding three-month period. Injuries included minor and major muscle strain with or without radiation into the back or shoulders, muscle spasm, sensory deficits in the distribution of the affected nerve root, and motor deficits with decreased deep tendon reflexes or impairment of coordination, dexterity, or movement. The high prevalence of (non-fracture) neck injuries in non-ejection situations would seem to lend support to the use of conservative measures of injury tolerance in design of emergency escape systems.
von Gierke, H. E.; Kaleps, I.; Brinkley, J. W.	1988	Human crashworthiness and crash load limits		fracture probability			3-axis					This paper describes in detail the acceleration exposure limit method used by Brinkley, Specker, and Mosher in their 1989 reference. This method makes use of a one-mass model for three-axis, whole-body dynamic response estimation and prediction of injury probability in pilot ejections and other impact scenarios. An Injury Potential Function is described. This quantity has a different value at each vertebral level and is obtained by dividing the maximum predicted stress at each level by the corresponding vertebral level mean failure stress. The Injury Potential Function predicts the observed result of higher probability of injury in the middle thoracic region of the spine than in the lumbar region in the case of very tight torso restraint. Injury potential (probability) as as determined from a head-spine model is graphed as a function of vertebral level for +Gz equal to 14, 16, 18, and 20 G.
Patrick, L. M.	1987	Neck injury incidence, mechanisms and protection	X		X							This paper includes a thorough discussion of the anatomy of the human neck, the incidence of neck injuries in the U.S., injury mechanisms and resulting injuries, and neck tolerance to dynamic loading. The author references experiments with human volunteers in which voluntary tolerances were determined as follows: forward head/neck flexion torque, 65 ft-lb; extension torque, 22.5 ft-lb; lateral flexion torque, 33.3 ft-lb; A-P or P-A shear force, 190 lb; R-L or L-R shear force, 90 lb. (All values are determined at the occipital condyles.) Marginal injury level exposures for cadaver subjects were determined as follows: forward head/neck flexion, 140 ft-lb (no damage); extension, 35 ft-lb (no damage); extension, 42 ft-lb (damage to ligaments). The author hypothesizes that, on the basis of these data, it may be true that marginal injury levels (to cadavers) are about double the voluntary tolerances. The voluntary axial force for the neck in tension or compression given as 250 lb.
Bosio, A. C.; Bowman, B. M.	1986	Simulation of head-neck dynamic response in -Gx and +Gy						bending and axial stiffnesses, damping coeffici- ents, and energy restitu- tion coeffici-				The study reported is a follow-on to the study described below in Bowman, et al., 1984. The goal of this study was to establish improved values for mechanical constants pertinent to the design of a neck for a Biofidelic Manikin. Values established in the first study were mostly unchanged in this second study. The primary changes were nonlinearization of the flexion bending stiffnesses for the base of the neck (C7/T1) and the tensile axial stiffness of the neck. Additionally, the authors recommend placement of the base of the manikin neck at a location 4 to 5 cm rearward and 3 to 6

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Bosio and Bowman, 1986 (continued)								ents				cm downward from the location of anatomical C7/T1 rather than at the anatomical location. This was found important for proper replication of human subject, relative angles in the neck by computer simulations (or equivalent manikin tests). Without this adjustment in representation of a two-joint neck it is possible to replicate the full head-to-torso angle response while predicting the head-to-neck and neck-to-torso relative angles incorrectly. The full report on this study is the 1986 report by Bosio and Bowman (below).
Bosio, A. C.; Bowman, B. M.	1986	Analysis of head and neck dynamic response of the U.S. adult military population						bending and axial				(See Bosio and Bowman, 1986, above: "Simulation of Head-Neck Dynamic Response in -Gx and +Gy". Also see Bowman, et al., 1984, below: "Simulation Analysis of Head and Neck Dynamic Response".)
Coltman, J.W.; Van Ingen, C.; Selker, F.	1986	Crash-resistant crewseat limit-load optimization through dynamic testing with cadavers		based on compressive strengths for vertebral fracture								This study investigated the threshold of thoracic and lumbar spinal injury for seated humans subjected to +Gz loading. Tests were conducted with human cadavers and with a Part 572 dummy modified with a 6-axis load cell at the base of the lumbar spine. Accelerations used were representative of vertical accelerations in helicopter crashes, with maximums of about 15 G reaching the occupant through various tested energy absorber load-limiting seating systems. Conclusions included the following: a) Compression testing of thoracic and lumbar vertebral segments is a reasonably reliable indicator of spinal strength, and the primary parameter of interest is the ultimate failure load; b) in tests with the dummy the measured spinal loads showed a strong relationship to the energy absorber limit-load factor of the seating system and to the angle of the seat with respect to the acceleration vector; c) the ratio of spinal load to vertebral compressive strength is a good indicator of the potential for spinal injury. Vertebral compressive strength data from a number of sources are given. The authors find that in good approximation the ultimate compressive strengths of thoraco-lumbar vertebrae increase linearly, by level, from T1 to L5. (See Table 4 and Figure 30 in their report.) Where L=1 for T1, L=2 for T2, ..., and L=17 for L5, the average ultimate strength may be determined from their data as S = 335 + (L-1)*(2015-335)/16, where S is in pounds. The greatest ultimate strength for any of the cadavers is S = 1193 + (L-1)*(3881-1193)/16. The least-bound ultimate strength for the cadavers is S = 200 + (L-1)*(1400-200)/16. (The slopes for the average, greatest, and least-bound strengths are 105 lb/level, 168 lb/level, and 75 lb/level, respectively.) These data are from 12 cadavers. The ages ranged from 44 to 63 years (average, 56.25); eight of the 12 were male.
Hayes, C. D.; Wasserman, J. F.; Butler, B. P.	1986	Effects of helmet weight and center-of-gravity on the vibratory dynamics of the head-neck system: a modeling approach						gross data				The authors use a four-degree-of-freedom simulation model of a weighted helmet, head, and neck to study vibratory response. The frequency response of the system was determined for different helmet masses and for four different locations for attachment of avionics mass. The model would probably be more useful in relation to prediction of fatigue than for prediction of response in non-vibratory, unidirectional +Gz pilot ejections.
Hearon, B. F.; Brinkley, J.W.	1986	Effect of seat cushions on human response to +Gz impact		pertinent		pertinent to head and chest acceleration limits						Human response to +Gz impact acceleration was evaluated as a function of various seat cushions, including current operational cushions and proposed alternative cushions made with rate-dependent, slow-recovery polyurethane foams. One hundred thirty-three tests were conducted of volunteer subjects in seven different experimental conditions, using a vertical deceleration tower facility. All tests were at a nominal +10 Gz. Responses measured were head and chest accelerations, belt loads, and seat loads. Use of rate-dependent foam cushions was found to improve the impact protection performance of escape systems.
Belytschko, T.; Rencis, M.; Williams, J.	1985	Head-spine structure modeling: enhancements to secondary loading path model and validation of head-cervical spine model	pertinent		cervical			for discrete elements	for discrete elements			The SAM computer simulation model is described. SAM (for Structural Analysis of Man) is a three-dimensional, discrete element model used for prediction of the dynamic response of the head-spine-torso structure to severe impact environments. (SAM is also called the HSM model at AAMRL, which supported and assisted in its development.) SAM includes representation of the head, torso, pelvis, intervertebral discs, ligaments, muscle, and other connective tissues. The effects of muscle can be simulated with either a passive muscle model or a stretch reflex model. SAM incorporates a data base that contains biomechanical, geometric, and structural data. The model has been used successively to reproduce results of

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Author(s)	Date	Title	Cervical Spine Dislocations and Fractures	Thoracic, Lumbar, and Sacral Spine Dislocations & Fx	Musculature and Ligaments	Head	Other	Cervical Spine	Thoracic, Lumbar, and Sacral Spine	Head	Other	
Belytschko, et al., 1985 (continued)												human volunteer tests in -Gx and +Gy. Improved modeling of viscera is said to make the model more able to predict response to +Gz inputs, but no +Gz simulations are presented. The authors discuss spinal injuries associated with pilot ejections and describe mechanisms for numerous types of injury to the cervical spine.
Alem, N. M.; Nusholtz, G. S.; Melvin, J. W.	1984	Head and neck response to axial impacts	axial, for S-I head impacts					axial properties				Nineteen impacts to the head in the superior-inferior direction using unembalmed cadavers are reported. Some impacts were used to study sub-injurious response and to determine mechanical characteristics of the system. The 10-kg impactor produced cervical spine injuries for impact velocities between 7 and 11 m/s. Peak impact force was not found to be a reliable predictor of cervical injury, nor was HIC. Peak head velocity was the best indicator of injury of all response parameters measured. Of the impact parameters examined, the integral of the force-time curve was the most consistent indicator of cervical injury.
Bowman, B. M.; Schneider, L. W.; Lustick, L. S.; Anderson, W. R.; Thomas, D.J.	1984	Simulation analysis of head and neck dynamic response						bending and axial stiffnesses, damping coefficients, and energy restitution coefficients				The objectives of the study reported were to quantify the biomechanical properties of the human neck which govern head and neck dynamic response and to establish the mechanisms responsible for primary aspects of response. Computer simulations with two- and three-dimensional occupant dynamics models were performed using head and neck sled input response data from human subjects at the Naval Biodynamics Laboratory for input and comparison. Simulations were done for peak accelerations below and up to maximum voluntary levels for the NBDL subjects. All simulations were for -X, +Y, and -X+Y acceleration vectors. Since significant axial loading of the neck occurs, however, even for these (non-Z) vectors, the study was able to establish approximations of the most appropriate axial properties of the human neck. In particular, estimates were established for compressive (and tensile) stiffness, damping coefficient, and energy restitution coefficient. Results with smaller error bounds were determined for non-axial neck properties since head/neck response for the vectors simulated was more greatly dependent on non-axial properties. In particular bending stiffness, damping, and energy restitution properties were determined for flexion, extension, and lateral flexion at both the neck-head articulation (condyles) and the neck-torso articulation (C7/T1), and the same mechanical properties were determined for head rotation relative to the torso. All of these non-axial properties should be represented in a manikin neck that is subjected primarily to axial (+Gz, ejection) inputs since non-axial head/neck responses will always occur. Neck mechanical properties established in the reported study are too extensive to include here. The full report on this study is the 1982 report by Bowman and Schneider (below).
Goldsmith, W.; Ommaya, A. K.	1984	Head and neck injury criteria and tolerance levels	shear, tension, compression		cervical	HIC, MSC, angular		bending stiffness		A-P properties		This paper compiles and reviews head and neck injury criteria and tolerance level data from numerous sources. A variety of injury modes are considered including some that have direct pertinence to +Gz inputs.
Helleur, C.; Gracovetsky, S.; Farfan, H.	1984	Tolerance of the human cervical spine to high acceleration: a modelling approach	"joint stress" strength; +Gz limit		neck muscle and ligament flexion-tension strengths; +Gz limit							This study used a sagittal plane mathematical model of the cervical spine and upwardly acting acceleration vectors at various angles. It assumed that cervical injury thresholds are determined by the tension load-bearing capability of neck muscles and ligaments during flexion of the head/neck system or by "joint stress" if flexion is not large. Limits are given for the neck for compressive joint stress, ligament moment, and muscle tension load. Both ultimate limits (strengths) and voluntary limits are reported from a separate study. It is stated that voluntary limits are not greater than two-thirds of ultimate limits. A conclusion is made that accelerations of up to 40 G will not cause injury to cervical muscles and ligaments or vertebrae with appropriate crewman posture and direction from vertical of the acceleration vector.
King, A. I.	1984	The spine - its anatomy, kinematics, injury mechanisms and tolerance to impact							T1-to-pelvis bending stiffness			A detailed anatomical description of the human spinal column is given in this paper. Additionally, all major classifications of spinal injuries are described. Experiments are described in which cadavers and human subjects were tested for spinal column flexure in -Gx sled tests. Response angulation data for T1 with respect to T12 and T12 with respect to the pelvis are presented. It was found that there is greater freedom for flexion in the thoracic spine than in the lumbar spine. Torque stiffness for rotation of T1 with respect to the pelvis was determined in static spinal flexion

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King, 1984 (continued)												tests with volunteer subjects who resisted applied torques. Stiffnesses ranged from about 0.08 N-m/deg to 0.17 N-m/deg. No injury criteria are given in the paper.
Ommaya, A. K.	1984	The neck: classification, physiopathology and clinical outcome of injuries to the neck in motor vehicle accidents	pertinent									This paper is thorough in its discussion of clinical and physiopathologic aspects of injuries to the neck. Included are identification of stable and unstable vertebral and disc injuries and their mechanisms. The paper also discusses cervical strain and relation of pathology to symptoms. Symptoms and signs of musculo-skeletal damage, and neural damage are also discussed. The author stresses the importance of minimizing the degree of head impacts, which reduces both the potential for head injury and neck injury, and notes that serious neck injury seldom occurs in the absence of head contact. He notes, however, that the detrimental effects of the the less severe types of cervical injury are generally underestimated. No injury criteria were established in the study reported.
Reading, T. E.; Haley, J. L., Jr.; Sippo, A.C.; Licina, J.; Schopper, A. W.	1984	SPH-4 U.S. Army flight helmet performance, 1972-1983	pertinent			pertinent						The performance of the SPH-4 U.S. Army flight helmet was studied for aircrew involved in 112 helicopter accidents from 1972 to 1983. While the data base includes head and neck injuries suffered from crown impacts in the crashes, which might be related to injuries resulting from through-the-canopy ejections, the paper does not explicitly identify the relationship between crown impacts and specific types of injury. Thus, the data base probably contains information pertinent to trauma assessment for some ejection-related injuries, but the paper is not itself of direct usefulness in this regard. The average AIS value for (unidentified) injuries in which the most severe impact on the helmet was to the crown was 2.7. The authors judge that in 52 instances out of 208 increased energy absorption in the helmet liner would have reduced injuries.
Hodgson, V. R.; Thomas, L. M.	1983	The biomechanics of neck injury from direct impact to the head, experimental findings	axial compressive loading									Dynamic impact loading of the neck through direct head impacts of cadaveric subjects was studied with regard to numerous variables. These included impact location, line of action, energy level, concentrated vs. distributed loading, initial neck curvature, and protective gear. Among the most important stated findings is that for compression loading in general there are too many variables affecting cervical spine injury to publish suggested tolerance limits. Nonetheless, the authors present their "best estimate of axial compression tolerance for the adult population." Specifically, they estimate AIS equal to 5 for an axial compression force of 250 pounds or more for a duration of 30 ms or more, AIS equal to 5 for force greater than 850 - 20 x T pounds for T less than or equal to 30 ms, and AIS equal to 0 otherwise. With regard to compressive loading neck injuries that might be caused by ejection forces (not discussed in the article), results presented by the authors are consistent with a conclusion that cervical compression injuries--like thoraco-lumbar compression injuries--are more likely when the neck is flexed, i.e., when the ejectee is not seated erectly with head and buttocks back against the seat.
Kazarian, L.	1983	Classification of simple spinal column injuries	pertinent	pertinent								This article is comprehensive in its description and illustration of the anatomical, radiographic, and biomechanical aspects of common acute spinal trauma. The article is not directed toward spinal column injuries resulting from aircrew ejection but, rather, more generally to the spinal injury mechanics associated with aerospace, sports, and recreational activities. The author states that "at this time [1980, 1983], it is impossible to measure the forces involved in producing a particular injury mode either at the body-environment interfaces or within the body itself with any form of instruments. It is clear that spinal injuries are highly variable and complex with a number of vectors simultaneously playing a role in the mechanics of trauma. Adequate knowledge of the strength of the human spinal column to a particular exposure is required in order to identify the probability and severity of trauma." No quantitative injury tolerance data are given in the article.
McElhaney, J.; Roberts, V.; Paver, J.; Maxwell, M.	1983	Etiology of trauma to the cervical spine	ultimate strength	ultimate strength (in compression) as function of age				stiffness, full neck and elements				Dynamic force-deflection data are given for compression loading of the full cervical spine. Results are nonlinear (1200-3600 lb/in, 2400 lb/in net). Ultimate strength data are given for human vertebral cancellous bone as a function of age (approximately linear from 1000 psi at age 20 to 500 psi at age 80). The most important structural properties are stated to be load to failure, stiffness, energy to failure,

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McElhaney, et al., 1983 (continued)												and damping or energy absorbed. This article gives statistics for cervical fracture level in auto and motorcycle accidents for head impacts of different types. Frequency distributions center about the following levels: C1-2 for low facial impact (extension-tension), C5 for high facial impacts (extension-compression), and C5 for head impacts (yielding flexion-compression injury). Burst fractures of the C5 vertebral body are common and require 1400-1800 lb.
Phillips, C. A.; Petrofsky, J. S.	1983	Neck muscle loading and fatigue: systematic variation of headgear weight and center-of-gravity	pertinent		pertinent to cervical strain						neck muscle strength	A series of experiments was conducted to quantify the fatigue of neck muscles as measured by isometric endurance time after 30 minutes of side-to-side head rotation while wearing a weighted helmet. Six subjects were tested for 15 headgear configurations; there were five different centers of gravity for the helmet/avionics combinations and three different helmet/avionics weights. Isometric endurance tests at 70 percent of an unfatigued subject's maximum isometric neck strength established that there was no fatigue associated with three of the fifteen weight/location combinations, viz., low weight/forward-low location, low weight/lateral-low location, and high weight/aftward-low location. All other head loading configurations resulted in a significant reduction in endurance times. (The low, intermediate, and high weights tested were 3.2 lb, 5.0 lb, and 9.0 lb.) The study recommends minimal combined helmet and avionics weights, low centers of gravity, and attachment of equipment to the back of the helmet whenever possible. It may be reasonable to extrapolate from the results of this study to the conclusion that neck injury criteria for aircrew should be defined very conservatively with respect to critical loads and moments.
Bowman, B. M.; Schneider, L. W.	1982	Analysis of head and neck dynamic response of the U.S. adult military population						bending and axial				(See Bowman, et al., 1984, above. Note: The 1984 paper contains more detail in some regards than this 1982 report.)
Brinkley, J. W.; Hearon, B. F.; Raddin, J. H., Jr.; McGowan, L. A.; Powers, J. M.	1982	Vertical impact tests of a modified F/FB-111 crew seat to evaluate headrest position and restraint configuration effects	pertinent	pertinent	pertinent	pertinent						Vertical Deceleration Tower tests in +Gz were conducted with human volunteer subjects. Acceleration magnitudes were as large as 10.5 G in the 115 tests. The purpose of the program was to determine qualitatively the reductions (or increases) and redistribution of escape system loads acting on a pilot that result from changing the fore-aft headrest position, bracing hands on the knees, and using either a modified F/FB-111 crew restraint system or a double shoulder strap - lap belt restraint harness. It was found that compressive spinal column loads can be reduced by the pilot by bracing with hands on the knees prior to +Gz impact in comparison to hands-in-lap positioning. (Crossed-arms positioning was not tested.) It was found that a headrest location 2.25" forward of the plane of the seat back caused much more forward and downward head rotation, i.e., flexion, than a location 1" aft. Therefore, to minimize the likelihood of dangerously large cervical spine hyperflexion loadings, the more aft headrest position is indicated. The F/FB-111 harness was found to reduce potentially injurious loads on the pilot in comparison with the double shoulder strap - lap belt harness. All measures that were found to reduce the potential for injury from +Gz impacts increase loadings in X and/or Y, but in actual ejections (+Gz's greater than the 10.5 G of the tests) the increases would (probably) not increase the potential for associated injuries significantly. Regarding injury criteria, this report contains experimental data useful for relating escape conditions, but not forces and moments, to minor injuries, including: cervical pain, paresthesias, abrasions, contusions, and muscle strains. It also contains subject dynamic response data that might be useful in validation of a manikin developed for ejection seat testing.
Cheng, R.; Yang, K. H.; Levine, R. S.; King, A. I.; Morgan, R.	1982	Injuries to the cervical spine caused by a distributed frontal load to the chest	resultant axial/shear neck load in tension-flexion mode									The study described in this paper is not of great pertinence to ejection injury research since primary accelerations were in -Gx rather than +Gz, but some information in the paper is of interest. Six cadavers were tested in chest impacts of severity great enough to produce cervical fractures and fracture dislocations without head impact. The authors found in studies of auto accident data that neck injuries are rare when the head does not strike some part of the vehicle interior. On the basis of experimental results for a combined axial tension and flexion mode of inertial loading of the neck, the authors propose a neck fracture criterion of 1400 lb resultant neck load (vector sum of axial tension and

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Cheng, et al., 1982 (continued)												shear forces) at the base of the skull. The proposed value is described as a "conservative" indicator of probable fracture at the atlanto-occipital joint. It is stated that in the combined axial tension and flexion mode, the critical parameter governing injury is axial tension and that the role of shear and moment is unclear.
Desjardins, S. P.; Coltman, J. W.; Laananen, D. H.	1982	Development of improved criteria for energy-absorbing aircraft seats		Dynamic Response Index (DRI)								The focus of the reported study is development of guidance in the design of aircraft seats of improved crashworthiness. The pertinent loading vector is +Z as it is in aircrew member catapult ejection. Further, typical peak +Gz magnitudes, pulse durations, and Z-velocity changes are not dissimilar in survivable crashes and ejections so many of the results of the reported study are pertinent to Task 3 of the current study. The purposes of the reported study were to develop more rigorous and comprehensive design and evaluation criteria, to more completely understand the complex response of the human occupant and seating system in the crash environment, and to maximize the efficiency of such systems in providing crash protection to the occupant. The paper contains little information in the way of injury criteria although DRI's are calculated for many simulated crashes with both dummies and cadavers. The primary pertinence of this paper to Task 3 of the current study is indirect; in particular, results presented should be useful in anticipating and understanding the effects of changing system constants in ejection tests, such as dummy weight, for example. Specifically, the paper describes the effects of changing the following conditions independently of other conditions: a) input pulse shape; b) magnitude of input acceleration; c) velocity change; d) rate of onset of acceleration; e) dummy type; f) dummy percentile; g) cadavers vs. anthropomorphic dummies; h) seat energy absorber limit load; i) load-deflection characteristics of energy absorbers; j) movable seat weight; k) seat frame spring rate; l) seat cushion stiffness; and m) seat orientation.
Eppinger, R. H.	1982	Injury criteria and mathematical analogs for selected body areas	X		X	X	X					The author provides a summary of means for estimating injury severity that were considered most valid and accurate by the Biomechanics Group at NHTSA, DOT. The general objective is provide guidance in the estimation of injury severity on a continuous scale given measurable engineering parameters. The primary dependent variable used is the Abbreviated Injury Scale (AIS) Severity Code. The body areas addressed are the head, face, neck, thorax, abdomen, pelvis, femur, knee, and tibia. Of particular importance to prediction of injury in ejections (+Gz) are injury criteria for the cervical spine. Data cited are from Hodgson and Thomas (1980; 1981; 1983) and Mertz and Patrick (1972). No thoraco-lumbar spinal injury criteria are given.
Gracovetsky, S.; Farfan, H. F.; Helleur, C. D.	1982	Cervical spine analysis for ejection injury prediction	pertinent	pertinent	pertinent			muscle & ligament				A detailed sagittal plane mathematical model of the upper spine (C1 to T6) and skull was developed and used to determine the maximum acceleration "supportable" for the upper spine for different postures and various acceleration vectors. "Supportable," here, means two-thirds of the acceleration that produces a stress in a vertebra equal to its ultimate strength. (The two-thirds factor is not verified by the authors as appropriate, but instead was selected because they established in an experimental study that weightlifters will not voluntarily execute a lift that produces lumbar compression forces greater than two-thirds of ultimate strengths.) Acceleration vectors were in or approximately in the +Z direction, and acceleration time-history inputs were consistent with +Gz seat accelerations for aircrew ejections. The model is detailed with respect to vertebrae size and shape, overall spinal geometry, and muscle and ligament properties and their insertion and attachment points. The maximum supportable acceleration was found to depend on neck posture and orientation vis-a-vis the acceleration vector. The value determined for the worst case was 13 G (two-thirds of 20 G) and the value for the best case was 40 G (two-thirds of 60 G). The authors state that it is very important to minimize shear force components in the occiput/C1/C2 structure in order to promote safe pilot ejection. Additionally, their simulation results indicate that it is advantageous to externally stimulate appropriate neck muscles before ejection begins. They also believe that air bags would be beneficial in helping to maintain proper alignment of the spinal column. Numeric values are given for estimated voluntary limits for joint stress, ligament moment, and muscle tension.

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Privitzer, E.; Carroll, C. D.; Kaleps, I.	1982	Analysis of personal thermal control system (PTCS) on head-spine structure ejection dynamics		pertinent					values not included in report			Computer simulations of +Gz ejections were made with a Head-Spine Model (HSM). This is a discretized, three-dimensional model of the human head and torso, i.e., the inertial properties of the torso are apportioned to vertebral levels for corresponding torso cross sections. The vertebral levels interact through deformable elements representing spinal ligaments, articular facets, and intervertebral discs. To assess probabilities of fracture injury at separate levels of the spine from T1 to L5, the model calculates an injury criterion called the HSM Injury Function. This quantity represents the ratio, at each level of the spine, of the peak computed cortical shell compressive stress (due to combined axial compression and bending) to the ultimate compressive yield stress. The report does not give values for the ultimate yield stresses, and additionally the authors state that the HSM Injury Function definition and calculation are not validated.
Sances, A., Jr.; Myklebust, J.; Houterman, C.; Weber, R.; Lepkowski, J.; Cusick, J.; Larson, S.; Ewing, C.; Thomas, D.; Weiss, M.; Berger, M.; Jessop, M. E.; Saltzberg, B.	1982	Head and spine injuries	X	X								The authors describe experiments in which human cadaveric spinal columns were tested in axial loadings, to failure, in tension and compression and at low (0.13 cm/min) and high (100-142 cm/s) loading rates. Experiments were done with skulls plus isolated cervical spinal columns, with isolated thoraco-lumbar sections alone, and with intact torsos. A large amount and variety of experimental failure load data are given in the paper, but these data are probably of little value with respect to defining trauma criteria assessment for manikin ejections since most cadaveric subjects were much older than the average military pilot. The average age of cadaveric subjects was 62.1 years.
Kazarian, L.	1981	Injuries to the human spinal column: biomechanics and injury classification	pertinent									This paper has the same basic objectives and scope as the 1983 Kazarian reference. It has pertinence to the understanding of spinal column injuries resulting from aircraft ejections but does not deal directly with emergency ejection. Range of motion data are given for the cervical, thoracic, and lumbar-sacral regions of the human spinal column.
Naval Safety Center	1981	Aircrew Automated Escape Systems (AAES) data analysis program symposium, Volume I.		maximum ejection acceleration and rate of onset								There is a very limited amount of information about injury criteria related to +Gz ejections in this document. For ejections in which the spinal column is properly aligned, 25 G is given as an acceleration that can be supported without vertebral fractures. Regarding ejection through-the-canopy it is noted that there is increased risk of vertebral fracture but that fractures are not normally disabling. Short duration accelerations from "seat slap" may be 40 G or more in through-the-canopy ejections without concomitant injury. Rates of onset of +Gz acceleration as large as 500 G/s or more can be tolerated without injury if the ejectee is properly restrained and sitting erectly on a rigid, stable seat.
Naval Safety Center	1981	Aircrew Automated Escape Systems (AAES) data analysis program symposium, Volume II.	(The only pertinent information relates to manikin ejection test procedures and interpretation of test results.)									Deficiencies in head load data measurements in dummies (1981) used in through-the-canopy tests are discussed, but no data pertinent to assessment of potential for injury are presented. Deficiencies mentioned include: nonrepresentative compressive stiffness of the body of the dummy under inertial loading from +Gz acceleration; nonrepresentative restraint system constraint of the dummy immediately after head-canopy contact; misalignment of the helmet force transducer with the actual head-canopy force vector.
Naval Safety Center	1981	Aircrew Automated Escape Systems (AAES) data analysis program symposium, Volume III.	-- not of pertinence to Task 3 --									There is nothing of direct pertinence to the Trauma Assessment Criteria task of the current study in Volume III.
Naval Safety Center	1981	Aircrew Automated Escape Systems (AAES) data analysis program symposium, Volume IV.	-- not of pertinence to Task 3 --									There is nothing of direct pertinence to the Trauma Assessment Criteria task of the current study in Volume IV.
Reid, S. E.; Raviv, G.; Reid, S. E., Jr.	1981	Neck muscle resistance to head impact						elasticity and damping				The role of neck muscles in the head/neck flexion response to torso loading was studied with an analog model. Neck elasticity and damping coefficients were determined for a variety of conditions through simulation of the response of volunteer subjects. Among the factors considered were magnitude and kind of impact, anticipation of the impact by the subject, and the presence or absence of a preload. The elasticity coefficient of women tested was less than half that of the men. Reaction and contraction times for all subjects were about 25-30 ms. Neck muscle resistance was

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Reid, et al., 1981 (continued)												found to significantly affect the head/neck response for the levels of impacts studied.
Melvin, J. W.	1979	Human neck injury tolerance	failure in compression and in bending				larynx					The author states that because of the complex structural interactions which can occur between the components of the neck it is necessary to define injury criteria in terms of forces and moments acting on the neck, rather than the stresses and strains in the tissues that are actually damaged (e.g., the spinal cord, laryngeal cartilages, etc.). Failure data are cited in this paper from various studies. For direct loading of the larynx structural collapse occurs at about 100 lb. Fracture of cervical vertebrae occurs for compression loads of about 1280 lb. Failure in bending of the neck does not occur for flexion moments at the occipital condyles as large as 140 ft-lb or for extension moments as large as 42 ft-lb.
Culver, R. H.; Bender, M.; Melvin, J. W.	1978	Mechanisms, tolerances and responses obtained under dynamic superior-inferior head impact, a pilot study	failure loads									This is a report prepared by the Human Injury Criteria Task Force of the Human Biomechanics and Simulation Subcommittee Eleven superior-inferior impacts of unembalmed cadavers were performed to study the mechanisms, tolerances, and responses of the head and neck. A 9.9 kg padded impactor at 6.8 to 10.2 m/s velocity produced cervical vertebrae fractures with no basal skull fracture. The authors believe the primary mechanism to be compressive arching. A synopsis of the axial load injuries produced is given above in this table in the summary for the 1991 SAE Information Report, SAE J885 JUL86.
Kazarian, L. E.; von Gierke, H. E.	1978	The validation of biodynamic models		ultimate load (strength) data for T1 to L5								This paper is primarily a review of spinal column modeling done prior to 1978 and a discussion of methods for extending and extrapolating available animal and subinjury human data into injury ranges for humans. Data for ultimate breaking strengths for isolated human vertebral bodies between T1 and L5 are given as a function of spinal column level. Different results are obtained for fast and slow loading rates (0.889 m/s and 0.0000889 m/s, respectively), but for each rate the breaking strength increases linearly, by level, from T1 to L5. Where L=1 for T1, L=2 for T2, ..., and L=17 for L5, the strength for fast loading is S = 3200 + (L-1)*(14100-3200)/16, where S is in Newtons. For the slow loading rate the result is S = 2500 + (L-1)*(6400-2500)/16. (The slopes of these two lines are 681 N/level and 244 N/level, respectively.) The authors do not give information regarding age of the cadaver(s) from which vertebral specimens were taken or the number of cadavers used. Data presented by the authors from other researchers are bracketed by their data for fast and slow loading rates.
Zenobi, T. J.	1978	Development of an inflatable head/neck restraint system for ejection seats	pertinent		pertinent (cervical)	angular velocity and acceleration						An inflatable head/neck restraint system for ejection seats is described. The restraint device is a ring-shaped inflatable bladder that fits around the neck below the helmet. The primary purpose of the restraint is to limit forward head and neck rotation of the crew member during ejection (and also at the time of parachute opening shock). The author states that reducing the maximum head/neck flexion reduces the probabilities of cervical fractures and neck muscle strains, but no design goal for minimum allowed angle is mentioned. Nor were there any specific design goals with respect to maximum allowed angular velocity or acceleration, but the author notes that 30 rad/s and 1800 rad/s/s, respectively, are critical values.
Patrick, L. M.; Chou, C. C.	1976	Response of the human neck in flexion, extension and lateral flexion	lower bounds on injury-producing torques									Human volunteers participated in a study to determine maximum voluntary static neck torques for flexion, extension, and lateral flexion. Low level dynamic exposures were also studied. No cadavers were used in any tests so all data obtained are for subinjury ranges. All results fall within envelopes previously determined by Mertz and Patrick (1972) in tests with volunteer subjects and cadavers.
Tennyson, S.A.; King, A.I.	1976	A biodynamic model of the human spinal column	pertinent	pertinent								A human spinal column, including musculature, was modeled and used for simulation of response in +Gz impact accelerations. Peak +Gz was not greater than 6 G in any of the simulations. The model predicted that spinal musculature was incapable of affecting overall spinal column kinematics. However, as a result of muscle contraction, significantly higher local axial forces were predicted in the discs and facets than were predicted when muscle was absent. The authors believe that these forces could add significantly to the problem of spinal fractures in 20 G pilot ejections. Injury and injury criteria are otherwise not discussed in the paper.

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Bowman, B. M.	1975	Cervical injuries: frequency, etiology, and severity										The author summarizes findings from a review of 125 papers through 1975 as pertinent to frequency, etiology, and severity of clinically treated neck injuries. Most but not all resulted from automobile accidents. Except for minor ligament strain and neurological effects, vertebral fracture is the most common cervical injury. Any part of a vertebra may be fractured, but compression fractures of a vertebral body are common and are the most significant in terms of injury severity. Compression fractures are most common at C5 and C6 and are of two types, both of which result from hyperflexion: a) "wedge" fracture, in which the anterior portion of the body is crushed; and b) "burst" fracture, in which the body is shattered. Wedge fractures are usually not serious, but a very high incidence of cord damage accompanies burst fractures. This is the experience in ejection-caused compression fractures of C5 and C6 as well and suggests the need in manikin ejection tests for being able to measure, at least crudely, the distribution of compression forces in the lower neck as apart from a force only at, say, the anterior "lip" of lower neck "discs" in the manikin neck. No injury criteria data are noted in the report.
Payne, P. R.	1975	Spinal injury in the crash environment		Dynamic Response Index (DRI)					pertinent			The author describes the DRI model for estimation of probability of spinal injury of the seated human in +Gz environments (ejections or helicopter crashes). The DRI, or Dynamic Response Index, is calculated from the response of a simple mass-spring-damper model (one degree of freedom) of the seated human. The DRI is the square of the natural frequency of the system (i.e., the spinal stiffness k divided by the head-plus-torso mass) multiplied by the maximum compressive deflection that results from a +Z driving force or acceleration and divided by the acceleration of gravity. The DRI is thus nondimensional. The spinal injury rate as a function of DRI is presented in the article in semilog form. Some data given were calculated from tests with cadavers and some are from operational experience. A graph in the paper gives the following preliminary results for rate (r) as a function of DRI (based on operational experience): (DRI=13.3, r=.002) (DRI=14.9, r=.010), (DRI=16.8, r=.05) (DRI=19.4, r=.20) (DRI=21.3, r=.50) These results are the same as those presented by Brinkley and Shaffer in an earlier paper (1971). (See below.)
Rotondo, G.	1975	Spinal injury after ejection in jet pilots: mechanism, diagnosis, followup, and prevention		break load and associated G's								The author states that it is "general opinion" that lumbar vertebra break loads occur in ranges from 18 to 25 G. He states also that experimental measurements of minimum static loads causing fracture indicate 700 to 900 kg at T12 and L1.
Glaister, D. H.	1974	Evaluation of aircrew protective helmets worn during crashes and ejections				pertinent						Drop tests with head forms and Mk 2 and Mk 3 helmets were conducted to try to reproduce damage caused in 11 ejections, six crashes, and a bird strike. An attempt was made to correlate energy and impact force from the mechanical tests with clinical injuries to establish energies and forces in the actual ejections. Impact energy was found to correlate reasonably well with injury. Transmitted force did not. It can be expected, however, that the nature of the correlation relationship would be different for different kinds of helmets.
Prasad, P.; King, A. I.	1974	An experimentally validated dynamic model of the spine						X	X			This paper describes a simulation model of the human spine. Cadaver data are given for C1 to L5 spinal column vertebrae and disc masses, moments of inertia, and viscoelastic properties.
Chen, S. J.	1973	A mechanical model of the human ligamentous spine and its application to the pilot ejection problem							X			A discrete-element computer simulation model of the human ligamentous spine is described. The model was used to simulate pilot ejections for a low value for peak +Gz, viz., 10 G. Values are given for constituative properties of the discs, ligaments, and vertebrae. Simulation results predict that the largest compressive loads will be in the lower thoracic region of the spine and, specifically, at the anterior lips of the vertebrae. The author concludes from his simulations that the facet joints in the lumbar region and the ligaments in the upper thoracic region play an important role in preventing vertebral fracture injuries in those regions.
Ewing, C. L.; King, A. I.; Prasad, P.	1973	Structural considerations of the human vertebral column under +Gz impact acceleration		qualitative data of pertinence								A vertical accelerator was used in 75 tests in +Gz for 12 cadavers to confirm an hypothesis that a major cause of ejection vertebral fracture is the dynamic reaction of the vertebral column in the presence of improper restraint, i.e., that there are certain movements of the individual vertebral

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Ewing, et al., 1973 (continued)												bodies under +Gz acceleration that cause the characteristic ejection vertebral fracture. Only anterior-lip fractures occurred under the conditions of the tests. It was found that for all other conditions being the same, the number of fractures could be reduced by restraining the shoulders and pelvis to a rigid seat back and forcibly hyperextending the lumbar vertebral column in the area of L1 with a wooden block. Fracture criteria data in the report are probably not useful for manikin design because the average age of the cadavers was 60 years.
Prasad, P.; King, A. I.; Ewing, C. L.	1973	The role of articular facets during +Gz acceleration		pertinent								The authors describe +Gz experiments done with human cadaveric specimans instrumented with intervertebral load cells. It was shown that while vertebral bodies in the spinal column bear most of the load from +Gz inputs, the articular facets of both thoracic and lumbar vertebrae can also support a significant portion of the load--up to 50 percent if the spine is in hyperextension. This dual load path along the spine--vertebral bodies and articular facets--thus seems to make it possible to raise the fracture limit loads in the thoraco-lumbar spine by a considerable margin by putting the spine into a hyperextended mode. Contrarily, failure to limit flexion of the spine in pilot ejections will increase probabilities for anterior wedge fractures of the vertebral bodies.
Mertz, H. J.; Patrick, L. M.	1972	Strength and response of the human neck	lower bound and marginal-injury criteria for flexion and extension					torque-deflection loading and unloading				Human volunteers were subjected to static and dynamic environments which produced noninjurious neck responses in extension and flexion. Tests with cadavers were used to extend the data into the injury region. Moment, shear force, and axial compression force injury criteria are given. This paper is the source of much of the neck injury criteria data cited in the literature. The injury criteria determined in this study are summarized in Patrick (1987). (See above.) Torque-deflection loading curves for angulation of the head with respect to the torso are given. Loading/unloading curve envelopes are defined for both flexion and extension.
Portnoy, H. D.; McElhaney, J. H.; Melvin, J. W.; Croissant, P. D.	1972	Mechanism of cervical spine injury in auto accidents	pertinent									This paper discusses in clinical detail the mechanisms of cervical spine injuries that result from direct impact to the head. The data were obtained from auto accident patient examinations and records. No cases without both neck and head injuries were included in the study. Thus, the pertinence to ejection related cervical injuries is only for ejections in which neck injury results from head impact, i.e., most likely only in through-the-canopy ejections. While the paper does not give either cervical injury criteria or mechanical properties data for the neck, it has pertinence to interpretation of neck force and moment data and therefore to design for data collection in a test manikin. It was found that there are three primary mechanisms of neck injury each associated with a particular type of head injury. In general the following statements can be made: 1) Injuries to the face produce extension-tension fractures. 2) Injuries to the forehead and frontal regions of the head produce extension-compression vertebral body fractures; inferior facet fractures are also usually seen. 3) Injuries to the parieto-occipital regions produce flexion-compression fractures, mostly from C5 to C7, and also anterior dislocations. The first two described mechanisms seldom produce neurological deficit while the third generally does. It may be noted that parieto-occipital head impacts produce the same type of injuries as ejections without head contact but with ineffective upper torso restraint.
Band, E. G. U.	1971	Calculation of rocket powered trajectories of a "plane of symmetry" model of a human subject and ejection seat	pertinent	pertinent								A five-mass, plane-of-symmetry model of a human subject and ejection seat is presented. It was found from simulations with the model that the rate of onset of rocket thrust is a critical determinant of the magnitude of spinal loads. Increasing +Gz on the seat from 1 G to 12 G over 10 ms resulted in nearly double the maximum spinal compression load as for a constant 12 G acceleration. Compression forces in the thoraco-lumbar spine were three to four times as large as neck compression forces in the simulations. Simulated thoraco-lumbar compression force was about 1100 lb for the case of a 10 ms rise time to a constant 12 G acceleration (through 200 ms).
Band, E. G. U.	1971	The dynamics of an ejection seat catapult with a "live load"		pertinent								A five-mass, plane-of-symmetry model of a human subject and ejection seat is used to compare the spinal forces during ejection for a subject that is compliant in the Z direction with the Z-forces on rigid, 328-lb man/seat system subjected

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Band, 1971 (continued)												to the same catapult loading. Little difference was found in peak simulated spinal compression forces.
Brinkley, J. W.; Shaffer, J. T.	1971	Dynamic simulation techniques for the design of escape systems: current applications and future Air Force requirements		Dynamic Response Index (DRI)					pertinent			The paper describes the one-degree of freedom Spinal Injury Model used for determining exposure limits for short duration +Gz accelerations produced by ejection catapult. The probability of compression fracture of spinal vertebrae is predicted by the DRI response (Dynamic Response Index) for the mass-spring-damper system. The authors reference system constants determined by Stech and Payne from experimental data--specifically, 0.224 for the damping ratio and 52.9 rad/s for the natural frequency. The probability (i.e., rate) of spinal injury from operational experience has an approximate power relationship to DRI. A graph in the paper gives the following approximate results for rate (r) as a function of DRI: (DRI=13.3, r=.002) (DRI=14.9, r=.010), (DRI=16.8, r=.05) (DRI=19.4, r=.20) (DRI=21.3, r=.50)
Leverett, S. D., Jr.	1971	An introduction to the physics and physiology of acceleration					vision, hearing, and other physiological responses					This article is Chapter II of AGARDograph No. 150. It includes physiological response data for exposure to varying levels of +Gz in the subinjury range. For the most part, then, the data here are not pertinent to the current task, Trauma Assessment Criteria, except perhaps as regards minor injuries of limited type.
Li, T. F.; Advani, S. H.; Lee, Y. C.	1971	The effect of initial curvature on the dynamic response of the spine to axial acceleration						constitutive properties, C1 to L4				A simple continuum representation of the spinal column was subjected to pilot ejection boost forces in computer simulations. Axial and lateral (bending) dynamic responses were studied. The numerical results indicated that the dynamic bending stress is significant in comparison to the axial dynamic stress. Values from the literature are presented for constitutive and geometric constants appropriate for the continuum spine model. It is noted that it is appropriate to use an elastic modulus larger than reported quasi-static values because strain-rate hardening is observed in most biological materials.
Nuttall, J. B.	1971	Emergency escape from aircraft and spacecraft		+Gz acceleration and rate of onset								The author gives a thorough discussion of factors related to proper design and deployment of ejection escape systems. A small amount of injury criteria data is included in the paper. He summarizes human tolerances to large-Gz environments in terms of approximate values or ranges as follows: +Gz, 20 G; -Gz (for downward ejection seat), 12 G; 250 G/s rate of onset,upward; 125 G/s rate of onset, downward; other values, +Gz of 25 G and rate of onset of 300 G/s. It is noted that accelerations to the required ejection velocity should be over at least 230 ms. The author notes that there have been accidental noninjury-producing exposures of human subjects to 30-33 G at 500 G/s rate of onset in upward ejection experiments under ideal laboratory conditions.
Orne, D.; Liu, Y. K.	1971	A mathematical model of spinal response to impact						X	X			This paper presents a discrete-parameter mathematical model of the human spine for simulation of pilot spinal column response in rocket-powered ejections. Cadaveric data are given for T1 to L5 vertebrae and disc masses, moments of inertia, and viscoelastic properties.
Payne, P. R.	1971	Some aspects of biodynamic modelling for aircraft escape system	compressive breaking strengths	compressive breaking strengths				axial compressive stiffnesses				This paper is largely a presentation and analysis of spinal column stiffness and strength data from various sources although it also includes some simple modeling work pertinent to aircrew ejection. Presented experimental data from young adult male cadavers include the following: compressive stiffness of a T11-to-L2 segment, 620 kg/cm; compressive stiffness of a T8-to-T12 segment, 1325 kg/cm; mean lumbar, single-vertebra stiffness, 15000 kg/cm; stiffness of lumbar disc (alone), 11500 kg/cm; vertebral stiffness, varies approximately linearly from 5080 kg/cm to 7400 kg/cm from T1 to T9; vertebral stiffness, varies approximately linearly from 7400 kg/cm to 11000 kg/cm from T9 to L5; vertebral strength, varies approximately linearly from 275 kg to 1000 kg from C4 to L1 (by level, for one adult male, age 30); vertebral strength, varies approximately linearly from 615 kg to 1067 kg from T8 to L5 (by level, for ten adults, age 19 to 46 with an average of 32.4). Data are also given for the mean vertebral body endplate area, by level, from L1 to L5. The author states that available data show that compressive breaking load is independent of age up to about 42 years and that it decreases exponentially above that. A curve that shows cumulative probability of compressive failure as a function of load is given. The data are normalized to L5 and age 42 years and are based on tests of 62 vertebral bodies. The article includes data that show that vertebral body

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Payne, 1971 (continued)												stiffnesses (K) are very nearly proportional to compressive strengths (S): S = 167.5 + .7793 K, where S is in kg and K is in kg/mm, where results are normalized to L5 and age 42 years.
Payne, P. R.; Shaffer, D. A.	1971	An optimum acceleration-time history for an escape system		pertinent					pertinent			A simulation study was conducted using a single-degree-of-freedom, idealized occupant and four different acceleration-time histories of the escape system. The occupant is represented with a mass for the head and upper torso and a spring plus damper for the spine (30% critical damping). It is noted that large ejection accelerations are required so that escape will be fast enough for clearance of the aircraft empennage. An upper bound to the acceleration-time history is imposed by the physiological limitations of the human pilot, i.e., by established injury criteria. The described study attempts to determine an optimum acceleration-time history which will, for a given physiological stress level, minimize the escape time and maximize the escape velocity. The authors conclude that an impulsive velocity spike, followed by a zero rise time constant acceleration, will give a 70% reduction in escape time relative to systems which employ a simple linear onset.
Sandover, J.	1971	Measurement of human responses during impact		velocity change for given peak acceleration and duration					natural frequency			Experiments are described in which human volunteer subjects were subjected to +Gz impacts similar to those in jet aircraft ejections. Pulse durations were 200 to 250 ms and peak accelerations were 5 to 6 G's. It was found that the resonant frequency of sitting human subjects (for +Gz excitation) was about 10 Hz. Regarding tolerance limits the author cites other studies which determined a 50 percent probability of injury if the total velocity change in ejection is 6 m/s.
Shannon, R. H.	1971	Operational aspects of forces on man during ejection/extraction escape in the US Air Force 1 Jan 1968 - 31 Dec 1970		+Gz acceleration, duration, and rate of onset								The author primarily discusses injury rates in U.S. Air Force ejections (1968-1970), but some comment is made regarding injury tolerance limits. The author cites as conservative maximum limits for +Gz the following criteria: 25 G for a duration of 100 to 150 ms and a 500 G/s rate of onset.
Symeonides, P. P.	1971	Some observations on compression fractures of the spine in ejected Greek pilots		pertinent								The author states the personal opinion that a pilot may resume unrestricted flying without increased injury risk after sustaining a wedge compression fracture in an ejection provided that the wedging does not exceed one third of the height of the vertebral body.
Terry, C. T.; Roberts, V. L.	1968	Viscoelastic model of the human spine subjected to +gz accelerations						overall spinal stiffness and damping constituative properties for a continuum representation				The authors present a continuum model of the human spinal column for use in +Gz simulations. The model is described as being an improvement of the earlier model of Hess and Lombard in that it adds damping in a uniform rod of Maxwell-type (series) viscoelastic medium. A best fit of experimental and simulation results was obtained for a modulus of elasticity of E = 2.5e4 lb/ft/ft and a viscosity of 1/mu, where mu = 3.75e-3 ft*ft/lb/s. While endpoint-to-endpoint acceleration response of the model was good, the authors state that this model is not suitable for prediction of injury-producing stresses along the spinal column. They believe a Kelvin model for a discrete number of vertebral elements would be more suitable.
Henzel, J. H.	1967	The human spinal column and upward ejection acceleration: an appraisal of biodynamic implications		vertebral breaking strengths in compressive loading								This report includes thorough discussion of spinal column anatomy, the physiology of the vertebral column, spinal biodynamics during ejection acceleration, and the pathogenesis of ejection spinal fracture. Breaking strength data are given for thoraco-lumbar vertebra (T1 to L5) subjected to compressive loading. The strength given for T1 is 360 lb and strengths increase linearly by vertebral level through 840 lb at T5. Strengths increase approximately linearly by level from 1000 lb at T6 through 1632 lb at T10. Strengths increase slowly between T11 and L1--1700 lb to 1790 lb. Strengths increase further from L2 to L5, again in an approximately linear fashion, from 1925 lb to 2366 lb. The strength data cited by the author are from Ruff (1950) and Stech (1963). All data here are for young, adult males.
Higgins, L. S.; Enfield, S. A.; Marshall, R. J.	1965	Studies on vertebral injuries sustained during aircrew ejection		ultimate compressive strengths for vertebrae and discs					compressive stiffnesses for discs			A literature study was conducted for the primary purpose of defining a research program for determination of dynamic strength of isolated vertebral specimans from human cadavers. Such research done prior to 1965 was reviewed and suggestions were made for continued research. The ultimate compressive strengths of individual vertebrae as determined by Ruff (1950) for young, adult males are cited. (The Ruff values are also cited by Henzel. See Henzel, 1967, in this table.) Experimental data due to Brown, et al. (1957) are

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Higgins, et al., 1965 (continued)												also given. These are for the ultimate compressive strength and the elastic properties of the intervertebral discs. The average strengths for three cadavers were about 1100 lb at L2/L3 and 1250 lb at L5/S1 with a linear variation by level in between. Disc axial stiffnesses averaged about 16,200 lb/in in the L2 to S1 region of the spine.
Carter, R. L.	1959	Human tolerance to automatic positioning and restraint systems for supersonic escape			X	X	X					This paper discusses the A3J-1 Supersonic Escape System and the forces imposed upon the pilot by it. The A3J-1 automatically positions the pilot prior to ejection. The effects of seat bottoming, leg positioning and restraint, the automatic-positioning inertia reel, and arm retention were examined in specially contrived tests using volunteer subjects. It was determined that for conditions mimicking the conditions of actuation of the A3J-1 system, none of the pilot/escape system interactions (not including boost forces) were deleterious.
Watts, D. T.; Mendelson, E. S.; Kornfield, A. T.	1947	Human tolerance to accelerations applied from seat to head during ejection seat tests		at least 20 G for a velocity change of 60 ft/s								Ejection tests were conducted with volunteer subjects to try to learn more about human tolerance to +Gz acceleration. Peak accelerations in most tests were about 20 G. Durations were about 300 ms, catapult strokes were 40 to 60 inches, and end-of-stroke velocities were up to about 60 ft/s. Maximum rates of onset for acceleration pulses were 150 to 280 G/s. It was found that 18 to 21 G was tolerated repeatedly without injury, but the authors reach no conclusions as to maximum +Gz that can be tolerated under operational conditions. The authors note that in German tests, researchers concluded that 20 G is the limit of tolerance for an average crew member, and on the basis of the resistance of fracture of isolated vertebrae and the load carried by each vertebra, the Germans concluded that fractures in the lumbar region will not occur until accelerations reach 22 to 25 G.
Watts, D. T.; Mendelson, E. S.; Poppen, J. R.	1947	Laboratory test of aviator's ejection seat		greater than 22 G for a velocity change of 60 ft/s								Ejection tests with 14 volunteer subjects are described. The authors believe that 20 to 22 G is the "practical upper limit" for seat ejection experiments with living human subjects. (Also see Watts, Mendelson, and Kornfield, 1947, above.)

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