CRASH SURVIVAL AND INJURY PREVENTION IN AIRCRAFT ACCIDENTS

BY

SQUADRON LEADER Matthew E. LEWIS MSc, MB BCh, DAvMed, RAF
(RAF Centre of Aviation Medicine – RAF Henlow)

This is an edited version of the article that first appeared in the Journal of Defence Science, *Volume 6 Number 3, September 2001.*

ABSTRACT

Developing effective programmes for reducing injuries in aircraft accidents depends on gaining an understanding of how accidents cause injuries, the nature of the forces contributing to the injuries, and the characteristics of the types of accident under investigation. From the beginnings of manned flight, a continuing aim has been to protect occupants in aircraft crashes from the effects of impact. The author examines the history of aircraft accident investigation and, from a purely military perspective, outlines the development of escape systems. Determining human tolerance characteristics and escape acceleration limits, supported by anthropomorphic test dummies and mathematical models, highlight the sketchy nature of present databases. These factors, together with the medical investigation and the interpretation of injuries sustained in accidents, are crucial to developing adequate predictive models and in turn will aim to improve crashworthiness, crash survivability and in-flight escape systems.

Introduction

Aircraft crashes are uncommon but often serious, and survivability is of major concern in both military and civil aviation. In survivable accidents, improvements in aircraft crashworthiness (the ability of a aircraft structure to withstand a crash), design criteria, personal protective equipment and aircraft escape systems may make injuries preventable. Injuries can be produced in many ways and are not simply related to the peak impact acceleration. While the probability of injury and death clearly increases with the severity of impact acceleration, there is no single threshold of impact stress below which nobody is hurt and above which everybody is injured.

Accident investigations have to consider all aspects of the crash, but two issues are central. The first is the cause of the accident and is crucial to preventing similar accidents recurring. The second concerns the consequences of the accident and specifically, from an aviation medicine perspective, the injuries sustained by the aircraft's occupants. Any effort to improve in-flight escape systems and provide protection against crash injury requires not only a knowledge of the environment to which an occupant may be exposed in a crash or ejection but also an understanding of how much force a human can be expected to withstand in a given situation.[1] Personnel involved in an aviation medicine investigation of an aircraft accident should have an understanding of the basic principles of crash survivability and aim to establish:

- What were the causes of the injuries or fatalities?
- What specific interactions between the victims and aircraft structures/components resulted in the injuries or fatalities?
- If the aircraft had provision for in-flight escape, why did the victims fail to escape?
- Did the in-flight escape system contribute to the causes of the injuries or fatalities?

- If the fatalities survived the deceleration forces of the impact, why did they fail to escape from the lethal environment of the wreckage?
- Did pilot incapacitation or physiological aberrations cause or contribute to the crash?
- What changes could be made to prevent the injuries or fatalities from occurring if an accident were to happen again under similar circumstances?

This article examines the history of aircraft accident investigations from the early work carried out in the fledgling aviation industry and onwards and, for the purely military perspective, outlines the development of escape systems. Acceleration aspects are then discussed with respect to human tolerance characteristics, anthropomorphic test dummies, mathematical models and escape acceleration limits. Injury analysis is significant in the design of protection systems and highlights the sketchy nature of present databases, the further investigation of which is crucial to developing adequate predictive models and in turn to improvements in crashworthiness, crash survivability and in-flight escape systems.

HISTORICAL BACKGROUND

Aircraft impacts and the development of the principles of crashworthiness

The first flight of a powered aircraft took place in 1903; five years later, the first aviation fatality occurred. Orville WRIGHT was demonstrating his WRIGHT FLYER to the US Army at Fort Meyer in Virginia and an Army Lieutenant, Thomas SELFRIDGE, who himself was a pioneer aircraft designer, had volunteered to act as an in-flight observer. During the flight, the starboard propeller fractured and Wright lost control of the aircraft. It nose dived into the ground from a height of 75 ft. WRIGHT, although injured, survived the crash but SELFRIDGE sustained a fatal compound fracture of the base of the skull. This accident should have promoted the use of helmets for head protection, and a few pilots adapted helmets designed for other sports for flying, but the practice was not widespread.

Many descriptions and reports of the accidents occurring at this time commented on the lack of restraint harnesses. However, the perceived need for restraint harnesses did not initially come from the point of view of impact protection but from the need to retain the pilot within the seat during aerobatic manoeuvres. As a result, early harnesses were modified safety belts originally designed to prevent construction workers falling from high buildings.

In the years before World War 1, restraint harnesses were becoming increasingly available but many aviators were still reluctant to use them. Their concerns were that the belts would hold the pilot in the seat if the aeroplane turned over in a crash and thereby crush the pilot, or, if the belt could not be released rapidly, it could trap a pilot in the burning wreckage. A Royal Navy Medical Services report from 1915 suggested that safety belts should be provided for in-flight use but should be undone before landing.[2] This ill-founded advice, if followed, would have exposed the aircrew to greater risk of injury during one of the most critical stages of the flight, the landing. Fortunately, the report's author had changed his opinion by the following year when he wrote that:

> "Seat belts should be attached to the airframe and not just the seat so that the belt could restrain the pilot during a crash."[3]

By 1918, the importance of adequate impact restraint was being recognized, and the *General Rules and Regulation Governing Flying* stated that all aircraft must be equipped with safety belts for the pilot and passenger. Moreover, in the event of an accident, the belt should not be released until after the accident, as it would probably save injury, especially if the machine turned over.

Following World War 1, an analysis of military aircraft losses showed that over 90% of fatal aircraft crashes were due to " failure of the flyer himself". This finding led to programmes to reduce accident rates by improving aircrew selection, training and medical monitoring, and the development of the concept of airworthiness to reduce losses from engine or airframe failures. Improvements towards injury prevention were also made and, in particular, observations made by military medical officers indicated that fatal head injuries were caused by pilots impacting with the aircraft cowl. Subsequently, cutting away the cowl to give 8 inches of head clearance almost completely eliminated the incidence of these injuries.

An accident in 1917 turned out to be of particular significance for crash injury research. It involved a cadet in the Canadian Royal Flying Corps, Hugh DE HAVEN. His aircraft collided with another aircraft in mid-air and the ensuing crash resulted in his sustaining two broken legs, minor lacerations and bruises, and a ruptured liver, pancreas and gallbladder. In his own analysis of his accident, he concluded that his safety harness, which was described as being 5-6 inches in diameter with a narrow pointed buckle in the centre, and had been fitted around his waist, in part caused his injuries. Working for the Canadian military, he continued to attempt to explain the causes of injuries in survivable accidents. His assessment that better engineering and design of aircraft could prevent injuries was met with limited enthusiasm. He often came up against attitudes such as that flying was dangerous and the best way of preventing injury was to stay on the ground. With the end of the war, and somewhat demoralized by the lack of progress towards the prevention of impact injuries, De HAVEN lost interest in crash protection, at least for a time.

Cruciform restraint straps were developed during the inter-war years but, as these restraint systems did not incorporate a lap belt or other form of lower torso restraint, the pilots were able to slip forward under the harness during a crash. This so-called 'submarining' movement tended to increase the likelihood of sustaining chest, upper abdomen and upper vertebral column injuries. The use of one such cruciform harness, the British Sutton Harness, was also noted to cause death by strangulation owing to the pilot submarining and trapping his neck in the V formed by the two shoulder straps and the central attachment buckle.

In 1936, the US Army investigated an air crash in which both occupants died of head injuries. The examination of the wreckage found that the structure containing the front and rear cockpits was intact, but that the centre of each instrument panel was damaged and bloodied from impacts by the pilots' heads. The investigation of this accident led to the recognition of the need for adequate upper torso restraint, and comparative impact tests of a lap belt only restraint system with a lap belt modified to include shoulder straps were instigated.[4] The results showed that the volunteer subjects, when restrained with the lap belt only harness, violently jack-knifed forward and only tolerated a deceleration of up to 8g but decelerations of 15g were tolerated, with no forward displacement of the upper body, when the modified harness was used. The prototype shoulder harness was further improved and then incorporated into the US aircraft used in World War II. One of the main developments was the addition of a single-point release mechanism so the lap and shoulder straps could be released simultaneously. In addition, the aft ends of the shoulder straps were fitted to the aircraft frame, through a spring-tensioned locking mechanism, so that the pilot could lean forward while in the harness but lock the harness for landing and take-off. Also proposed at that time was an inflated rubber seat-back for use in transport aircraft, which would protect and support the passenger in the seat behind. Although this idea was not immediately developed, the concept was undoubtedly a precursor to cockpit and automobile airbags.

After having being involved in a minor automobile accident in 1935, De HAVEN renewed his interest in crash protection. He realized that engineers were unaware how many people were killed or injured by engineering design flaws that could be corrected, so he urged the US Government to set up a programme where doctors, engineers and safety groups could work together to reduce crash injuries in aviation accidents. Despite initial set backs, he eventually secured the support of the Safety Bureau of the US Civil Aeronautics Board and the Committee on Aviation Medicine of the National Research Council. Their support led, in 1942, to the establishment of a joint Crash Injury Research Project centred at Cornell University in New York. The studies conducted ultimately formed the basis for many of the subsequent advances made in crash injury protection and, by the end of World War II, De HAVEN had analysed and amassed data on 185 aircraft accidents.[5] From his analyses he concluded that:

- In accidents where the cabin's structure was distorted yet remained intact, most serious injuries and deaths were caused by dangerous cabin installations.
- Crashes involving forces sufficient to cause partial collapse of the cabin structure were often survived without serious injury.
- The head was the first and most vital part of the body exposed to injury, and the probability of severe injury to the head, extremities and chest was increased by the failure of the safety belt assemblies or anchorage.
- Head injuries were caused by impacts against heavy instruments and solid instrument panels, insufficiently strong seat backs, and unsafe design of control wheels.
- The tolerance of crash forces by the human body had been grossly underestimated.
- If spin-stall dangers were lessened and safer cabin installations used, fatal or serious injuries should be rare in the types of accidents studied, except in extreme cases.

The results obtained by the Crash Injury Research Project were presented to aircraft manufacturers, but at first only a few indicated their intention to provide improved spin performance, cockpit structure, safety belts and instrument panels.

The fundamental principles of accident survival developed by early accident investigators remain unchanged to this day. Thus, to survive an impact:

- A container must be provided to surround the occupant, who should be held in place by a seat and restraint harness
- Accelerations of the container and seat occupant should be within tolerable levels.
- The occupant should be provided with personal protection equipment such as helmets.
- Account should be taken of post-crash factors such as fire or underwater escape (FIG.1).

However, a key means of improving impact survival is to remove aviators from the aircraft before the impact occurs. By bailing out of the aircraft, or by the use of ejection seats, the crash impact can be avoided. However, while both eliminate exposure to certain impact forces, additional risks are introduced, including:

- Ejection acceleration.
- Wind-blast.
- Altitude exposure.

- Parachute opening shock.
- Parachute landing.

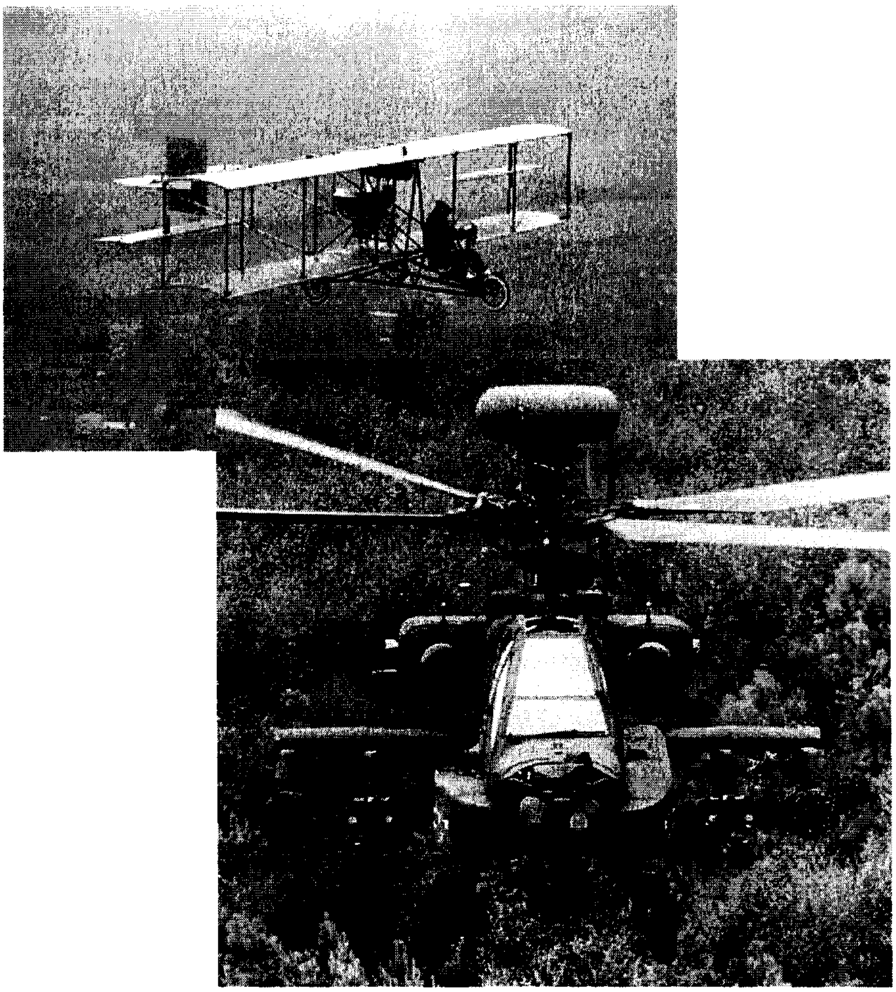

FIG.1 – EARLY AIRCRAFT PROVIDED LITTLE OCCUPANT CRASH PROTECTION WHEREAS, TODAYS AIRCRAFT HAVE A STRUCTURAL ENVELOPE SURROUNDING THE OCCUPIED AREAS TO MAXIMIZE ENERGY ABSORPTION. THE COCKPIT AND CABIN STRUCTURES ARE REINFORCED TO WITHSTAND LOADS INDUCED BY BLADE STRIKES, EXTERNAL OBJECTS AND ROLLOVER.

In-flight escape systems

The development of aircraft capable of flying at high speed resulted in the traditional method of leaving a disabled aircraft, by bailing out, becoming impossible. The wind-blast made it difficult to climb over the side of the cockpit, and it was especially hazardous if the aircraft was fitted with a pusher propeller sited aft of the cockpit. By 1940, a prototype ejection seat had been tested at the Heinkel Aircraft Facility in Germany. At first, these ejection seats, occupied by test dummies, were fired from an inclined ramp, but then human tests were carried out that culminated in BUSCH, a test parachutist, making the first successful manned ejection at a speed of 300 km h^{-1}.

The German development of the ejection seat raised questions concerning the tolerance of pilots to the acceleration loads imposed by the ejection seat. A thesis published in 1944 from the Technische Hochschule in Stuttgart presented an analytical discussion of the dynamics of ejection from aircraft, the development of a methodology to measure acceleration, and research on the biomechanics and impact tolerance on the spinal column.[6] The author recognized that the seat cushion's construction could influence the acceleration of the seat occupant and, depending on its composition, the seat cushion could reinforce or dampen the acceleration. In 1944, the success of the ejection seat trials led to a directive by the German Air Ministry that all fighter aircraft be provided with ejection seats. Heinkel ejection seats were duly fitted to the EH 162 jet fighter and Messerschmit 262; by the end of World War II, 60 ejections had taken place from Luftwaffe aircraft.

Similar studies were also being carried out in Great Britain and Sweden. At the RAF Physiological Laboratory (to become the Institute of Aviation Medicine in 1946), preliminary tests on rocket-propelled sleds attempted to define the limits of physiological tolerance to accelerations[7]. The sled produced an acceleration of 12g in the first 0.1 seconds for a total impact duration of 0.175 second. It was thought that the short acceleration distance of 6 ft over which the acceleration pulse took place was not representative of an ejection seat, so further tests were carried out using ejection seat towers. From the many live tests, it was concluded that:

- The acceleration acting along the spinal column should not be greater that 5g for the first 0.01 second of the impulse.
- A peak of 25g should not be exceeded at any stage in the ejection sequence.
- A rate of onset of acceleration should not be in excess of 300g per second.

In Great Britain, the first successful test dummy ejection took place, from a modified DEFIANT aircraft, on 11 May 1945 and, on the 26 June 1946, Bernie LYNCH, another test parachutist, ejected from a Gloucester METEOR fitted with a Martin Baker ejection seat at a speed of 515 km h^{-1}.

The early ejection seats were merely a means of separating the man from the aircraft, and the aircrew had first to jettison the hatch or canopy and then initiate the ejection, usually via a face blind. As the ejection seat had a high forward speed and a tendency to roll and tumble after it left the aircraft, the seat needed to be stabilized to allow the crewman to separate cleanly from the seat. This was achieved by a drogue parachute fired by a static line attached between the seat and the aircraft. Once the seat had stabilized, the crewman had to release his seat restraint harness and, when clear from the seat, pull the ripcord of the parachute. The four sequential manual actions required by the crewman to escape from the aircraft could be accomplished only with difficulty, and resulted in many mishandling incidents.

In some early systems, for instance in the CANBERRA, no pilot had succeeded in ejecting after the canopy had been manually jettisoned. But, in 1951 an American pilot successfully ejected from a CANBERRA having mishandled his ejection drill, and ejected inadvertently through the canopy. It was then thought that the reason why no successful ejection had occurred before this accident was the air blast to which the pilot was subjected after jettisoning the canopy. This could be sufficient either to prevent the pilot reaching, extracting and operating the face blind, or to dislodge the face blind so that it blew out of reach. Subsequently, the routine drill for CANBERRA pilots was to eject through the canopy and this has since been carried out successfully many times.

The high number of mishandling incidents led to the realization that fully automatic escape systems were required, and these came into RAF use in 1953 with the introduction of the Mk 2 ejection seat. Initiation of fully automatic seats via the face blind, or later via the seat-pan firing handle, started the sequence of events that resulted in the:

- Separation of the cockpit canopy from the aircraft.
- Ejection of the seat.
- Automatic separation of the seat at a pre-set barometric altitude after an acceleration and time-controlled delay.
- Automatic deployment of the parachute canopy.

Ejection systems have now been developed to a stage where either aircrew of a twin-seat aircraft can initiate the ejection despite the other crew member being totally unprepared for ejection. Thus the system has to pre-position the aircrew in the ejection seat by a harness retraction system while canopy jettison or canopy fragmentation devices are clearing the ejection path.

The introduction of the fully automatic escape system meant that the safe survivable ejection envelope was increased and survivable ejections were now possible from a minimum height of around 500 ft. Nevertheless, the development of faster aircraft, operating at lower levels, and with some being fitted with tall tail fins, called for ejection seats that could attain a greater height during ejection. The introduction of the Mk3 seat provided an increase in the end-of-gun velocity from 60 ft s^{-1} to 80 ft s^{-1}. An 80 ft s^{-1} end of gun velocity made a ground-level ejection a realistic possibility and was achieved successfully for the first time from a METEOR 7 aircraft in 1955. However, the ejection was only completed successfully because the aircraft had a forward velocity of 100 mph, which gave the seat an appropriate flight profile to allow the main parachute to deploy fully. Also, the vertical accelerations produced by ballistic ejection seats were now reaching the acceptable limits of human tolerance as shown by up to 70% of RAF ejectees sustaining spinal compression fractures on ejection.[8] To provide the capability for a zero altitude, zero speed ejection, and also a safe ejection from vertical take-off and landing aircraft (which potentially could have zero forward speed but a high vertical descent rate), the option was to develop a seat fitted with a rocket motor.

Rocket-assisted seats had been tested extensively in the late 1950s and early 1960s; in 1962, the first live ejection of a seat fitted with a rocket motor took place. This test proved the safety of the rocket-assisted seat. Thus rocket motors were fitted retrospectively to some earlier ballistic seats to form the Mk 6 and 7 seats. Rocket-assisted seats retain the cartridge-operated telescopic gun to provide the initial thrust but, as the seat accelerates upwards, at the end of gun travel, the rocket pack ignites and propels the seat further to increase the height gained before main parachute deployment. To obtain the high seat velocity needed to achieve a safe zero-zero ejection, the acceleration can act over a longer time, thereby limiting the peak acceleration and rate of onset of acceleration and, hence, the likelihood of spinal injuries (FIG.2).

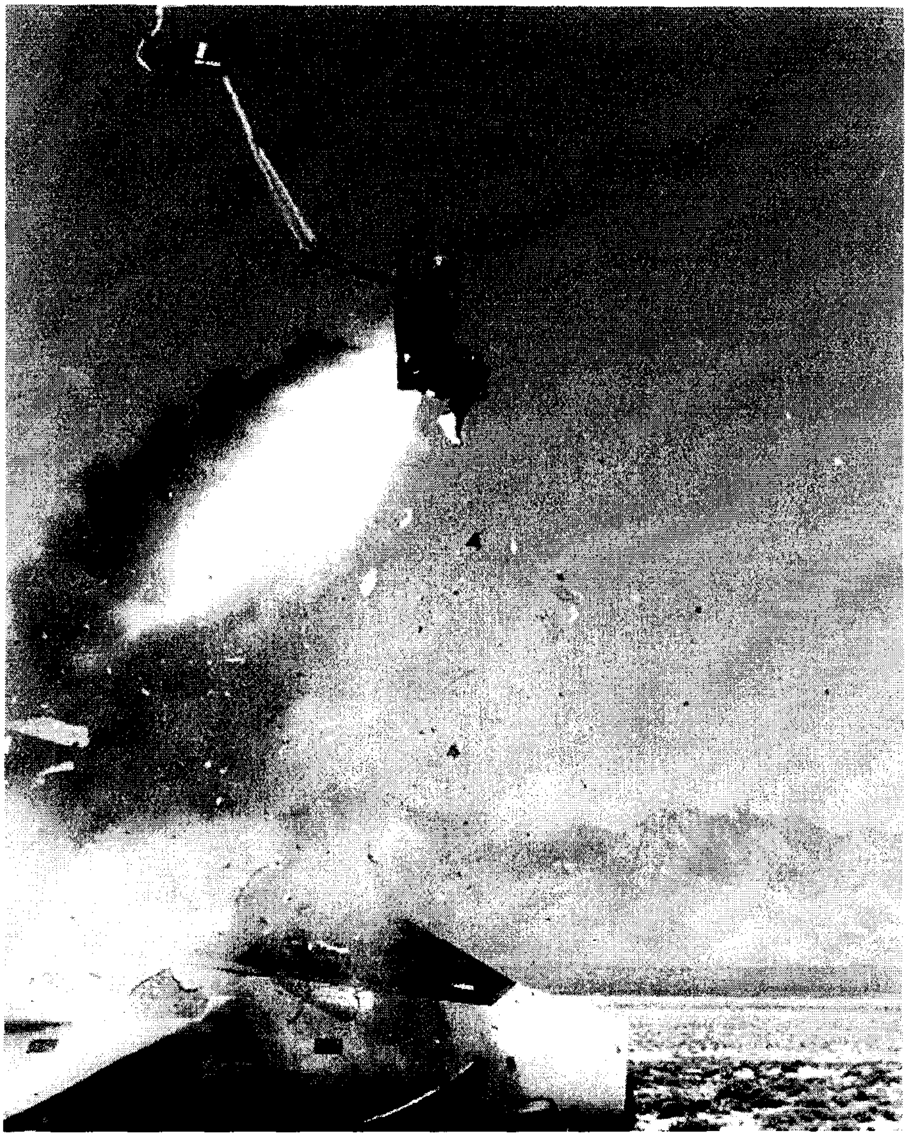

FIG.2 – HIGH SPEED SLED TEST OF A ROCKET ASSISTED EJECTION SEAT.

TOLERANCE TO ACCELERATION

Human characteristics and tolerance to impact

All approaches to designing emergency escape and crash protection systems are limited by the tolerance of the occupant to the acceleration and dynamic forces. Impact deceleration is characterized by forces of very abrupt onset, short duration and high magnitude; impact is conventionally defined in terms of forces lasting less than a second. However, such a limit is not applicable over the range of accelerations met in aviation, and it is more appropriate to distinguish long-duration and impact accelerations by the response of the human:

- Long-duration acceleration protection is relevant where stresses are primarily physiological and sustained.
- Impact acceleration protection when the stresses are primarily mechanical and transitory.

Overlaps do occur, however, and physiological events such as unconsciousness may arise from either, although the causative mechanisms differ. Unconsciousness following impact acceleration is caused by physical trauma, while with long-duration acceleration it is caused by reduced blood supply to the brain.

The current state of knowledge concerning human impact tolerances is incomplete. Most human volunteer studies have been conducted on young, healthy male subjects under rigidly controlled conditions with careful medical monitoring, and which have been terminated voluntarily at levels below those likely to cause irreversible injury. Little experimental data is available for females and, owing to the large range of human variability, data derived from experiments with volunteer male subjects must be used with caution. Animals have been used extensively to obtain physiological data levels above those potentially injurious to the human volunteers. Cadavers too offer a means of determining structural limits of tissues but they cannot provide the physiological information that can only be obtained from living systems. Determination of human tolerances to impact acceleration is complicated by many different physical factors including characteristics of harness restraint systems, body orientation, helmets and other equipment worn by the occupants, and the magnitude, direction, distribution, duration and pulse shape of the forces resulting from the impact. Human tolerances are not only influenced by gender but also by age, race, build, fitness and freedom from disease. Furthermore, tolerance under identical test conditions varies even in the same individual, as well as from person to person. Hence, attempts to quantify impact tolerance limits have resulted in approximations and generalizations making it necessary, in any one accident, to analyse causes of injury to occupants individually.

Anthropomorphic test dummies and mathematical models

The tolerance limits for fatality and injury have been derived from research carried out in a variety of institutions using a range of experimental devices and techniques. The limited numbers of impact assessments that can be conducted using animals, cadavers, and live volunteers, and the variability of the subjects themselves, have provided only approximate tolerance limits. The alternative, using anthropomorphic test dummies to provide repeatable impact conditions, has suffered from the many different types of anthropomorphic test dummy used, each with its own set of characteristic responses and limitations. The protocols, measurements and recording techniques used in these research programmes have been many and varied, making it extremely difficult to compare the results of tests using dissimilar anthropomorphic test dummies with results from tests using live subjects or cadavers. No experimental programme can fully reproduce the conditions met in an accident, and data from all experimental programmes require validation against injuries from painstakingly researched real accidents. Mathematical models are being developed to assist in the understanding of the nature of the forces encountered during accidental impact. Although these, and the new generations of anthropomorphic test dummies, are becoming more biofidelic, they are not human beings. Not only do mathematical models and anthropomorphic test dummies not break in an impact, they also lack the internal structure of the human body and so cannot realistically mimic the results of impact accelerations on organs and body tissues.

In-flight escape acceleration exposure limits

Linear and angular accelerations are significant hazards to aircrew safety during in-flight emergency escape. Nevertheless, owing to the scarcity of experimental data on the response of the body to multi-axial acceleration, criteria for design and evaluation of escape systems have been restricted to limits of acceleration, and rates of acceleration onset, for just vertical acceleration. Mathematical models have thus tended to be used to assess the probability of injury for accelerations acting in the g z-axis. Furthermore, as limits have not been set for angular acceleration, the mathematical models have been unable to incorporate angular acceleration data. Mathematical models, which are mechanical system analogues of the dynamic response characteristics of the human body, were developed to allow predictions to be made of the vertebral fracture rates from ejection seat test data. They were also to provide a method of estimating the acceptability of accelerations for an escape system, without having to wait for injury data to accrue. A number of complex models were initially explored, but simple single-degree-of-freedom, lumped parameter, models were deemed adequate to explain the limited available test data applicable to escape systems.

The model developed to the most satisfactory degree was the Dynamic Response Index (DRI),[9] which was aimed at estimating the probability of sustaining compression fractures in the lower spine from accelerations acting along the axis of the spine in the pelvis-head direction. The model was verified by comparing its response to ejection catapult accelerations with the operational injury rates associated with specific ejection systems. After operational verification and use of the model in the analysis of data from tests of developmental escape systems, it was incorporated into multinational specifications for ejection seats and escape systems.[10] The model, however, is not ideal and it makes some questionable assumptions:

- It is assumed that the spine can be represented mechanically as a simple mass/spring damper system with known natural frequency, and with damping characteristics that are invariant.
- The acceleration waveforms are assumed to be triangular or trapezoidal, and successive pulses are treated independently.
- The impact is assumed to occur in a healthy young male, seated on an inelastic cushion, adequately restrained by an inextensible harness, and wearing a protective helmet.

The data available to develop impact exposure-limit criteria are not extensive. Existing data from tests with volunteers at injurious as well as non-injurious levels are limited in the ranges of acceleration-vector directions and acceleration-time histories that have been explored. There is little information available on the effects of angular acceleration and combined angular and translational acceleration. However, the available human test data and the operational experience with emergency escape systems provide evidence that the current acceleration exposure criteria are insufficient. An analysis of injury data from RAF ejections has shown that the observed rates of spinal fractures are not well correlated with the rates predicted by the DRI.[8,11] The inadequacy of the DRI model for accurately predicting ejection spinal injury rates has demonstrated the need for better mathematical models.

A six degrees of freedom impact-evaluation model has been developed that has the advantage that the linear accelerations from the orthogonal seat axes are considered rather than just the Z axis, and neither angular accelerations nor angular rates are limited.[12] The model is based on available human test data together with operational escape experience, judgements founded on observations of the responses of volunteers to non-injurious acceleration conditions, and insights gained from computational simulations of human responses. The model provides a more comprehensive approach to evaluating the acceptability of whole-body accelerations so that the feasibility of new escape systems, and other personnel protection systems, can be demonstrated.

Injury analysis

Determining the sequence of events in an accident can provide information for injury-pattern analysis by relating various combinations of injuries, and certain characteristic patterns, to the sequence of events. In general, the types of injuries seen in aircraft accidents can differ markedly between aircraft types.. For instance, injuries obtained in a helicopter accident are different from those in a fast jet accident. Moreover, within a single accident, there may be dissimilar injury patterns amongst those individuals involved. The nature of the injuries may be explained by:

- Understanding the role of the crew member.
- His or her seating position at the time of the accident,

Analysing the injuries sustained can enable the dynamic forces involved to be estimated, although it is important to base this estimation of force solely on deceleration injuries and not on contact injuries (table 1).

TABLE 1. – *The approximate accelerations at which deceleration injuries to various body tissues and structures can occur.*[13]

Deceleration injury	**Acceleration (g)**
Compression fractures of the vertebra	20-30
Intimal tears of the aorta	50
Transection of the aorta	80 -100
Pelvis fracture	100 - 200
Transection of the vertebral body	200 - 300
Total body fragmentation	> 350

Head Injury

The head and neck are especially susceptible to injury in transportation accidents, with head injuries being the most frequent cause of death in aircraft accidents.[14] Death often results from the head and upper torso flexing forward and the head striking the instrument panel. Preventative measures such as helmets and shoulder restraints have helped to reduce, if not entirely prevent, these injuries. But, even with a torso restraint system in place, the head can still strike the instrument panel as a result of deformation of the fuselage or a cockpit strike envelope that is too small. The force of the impact can dislodge the helmet from the head, and injury may result. However, a fatal head injury can also occur with the protective helmet remaining in place and intact. The helmet may distribute impact forces widely over the head, leaving the scalp and skull undamaged, while fatal forces are transmitted to the brain. The pattern of skull fractures can give an indication of the direction of the applied forces. Linear fractures of the skull are most often found in the plane in which the force was applied. But this is not always the case, as forces from an impact to the chin may be transmitted through the arch of the jaw to the tempromandibular joints, causing a basilar skull fracture through the middle cranial fossa. Forces transmitted directly up the spine in a g-z impact can cause ring fractures around the circumference of the foramen magnum. Head injuries can sometimes mislead investigators if too great a reliance is placed on the examination of the strike envelope or collapsed fuselage, and the possibility of impact with loose cabin objects is ignored.

Thorax and abdomen

Crash forces can damage the internal organs of the chest and abdomen in several ways, including direct blunt or penetrative trauma, which may result in severe or fatal injuries. However, as the internal organs are relatively unrestrained, and are only suspended by other soft tissues and ligaments, they may experience tortional and shearing forces during the impact. The deceleration forces acting on these organs may be much greater than those acting on more restrained body parts, and as a consequence internal tears may result. The heart and great blood vessels may be compressed between the sternum and vertebrae and lead to rupture. Furthermore, their rupture may occur following a compression force to the chest or abdomen that transmits hydrostatic pressure backwards toward the heart. The application of forces by a restraint belt may produce rib fractures, the broken, jagged ends of which may lacerate the heart, great vessels and lungs, leading to fatal haemorrhage.

Injuries to limbs

Flailing of the arm and legs may lead to injuries of the limbs by impact with aircraft structures within the strike envelope. Although flail injuries are of limited use in estimating the impact velocity and parameters of the crash pulse, they may nevertheless produce patterns of injury that may indicate to the investigator exactly what structure the occupant struck. Injury patterns of the hands and feet may be useful in establishing who was in control of the aircraft at the time of an accident, or even if a solo pilot actually had his hands on the controls at the time of impact. Fractures of the hands may occur in those who were tightly holding the wheel or stick during the crash, and energy transmitted through the pedals may cause characteristic fractures of the feet.

Spinal injuries

Valuable information can be gained by examining the spine as data relating to the direction and magnitude of the impact can be obtained. Vertebral compression fractures can result from forces acting in the z direction and occur especially in impacts with a high vertical acceleration, such as hard helicopter landings or ejection from fast jet aircraft (FIG.3). Vertebral fractures tend to occur with vertical forces of around 20g, but fractures can occur at much lower g levels, particularly when positioning of the spine is not entirely vertical and the torso flexes forwards. Aircraft occupants who are only restrained by a lap belt can be at risk of sustaining characteristic injuries, which have been described as the 'lap belt syndrome'. In impacts with a high g-x deceleration, a lap belt can displace the fulcrum of the force anteriorly. The entire spine is then subjected to tension stress. This can result in disruption of the ligaments of the posterior elements of the spine or, should the ligaments remain intact, a transverse fracture of the posterior vertebrae may occur. Although rare, if such a transverse fracture is identified in casualties restrained only with a lap belt, associated abdominal visceral injury is also frequently seen, and 15% of cases have been shown to have an associated neurological deficit. The simple vertebral wedge compression fractures, which are frequently seen following ejection, are in contrast rarely associated with neurological deficit.

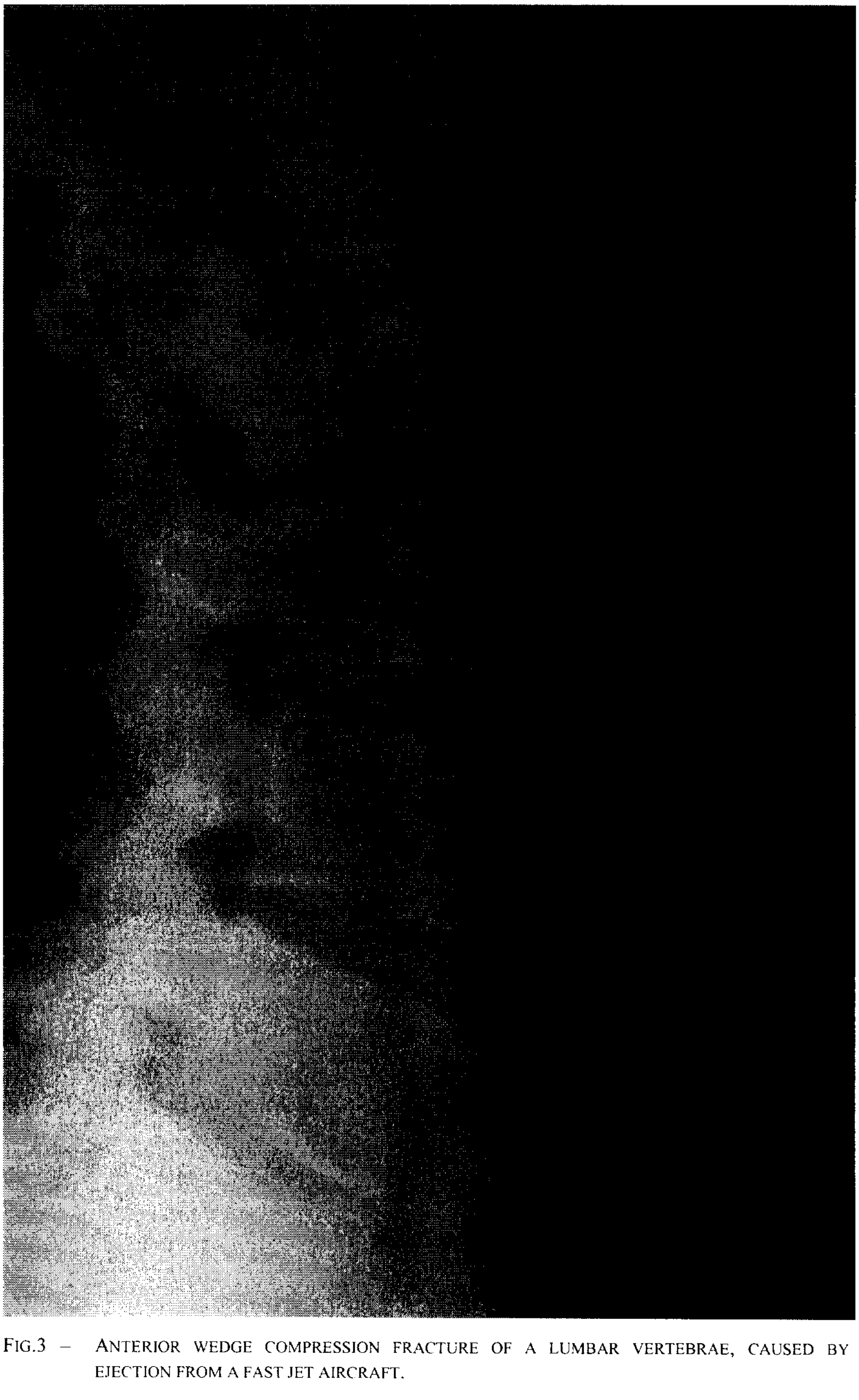

FIG.3 – ANTERIOR WEDGE COMPRESSION FRACTURE OF A LUMBAR VERTEBRAE, CAUSED BY EJECTION FROM A FAST JET AIRCRAFT.

Summary

The investigation of an aircraft accident in which injury has occurred must ultimately address what could be done to prevent similar injuries in the future. Recommendations for changes to improve crash survivability may simultaneously introduce other risks that would outweigh any advantages. Beneficial measures introduced in response to a specific impact may be harmful in another, and may pose additional problems during normal operations. Any protection system represents a compromise between a number of factors including weight, comfort, mobility, and occupant anthropometry. Gauging the success of a crashworthy design is difficult, as it is not always reasonable to assume that the occurrence of an injury in an impact implies a deficiency in the protection system. Any practical system can be exposed to an impact so severe that it is beyond its ability to provided effective protection – after all, no realistic crashworthy design can offer protection for Mach 1 impact into the side of a mountain! However, protection systems must be designed for the entire range of normal and emergency operations, and for the entire range of likely impacts. Unfortunately, the dearth of data, and the inadequacy of current impact models, make designing the ideal protection system an unrealistic proposition for the foreseeable future. There is an urgent need to enhance existing databases, particularly with regard to multi-axial acceleration injuries, so that models with greater predictive validity can be derived. Only then can new protection systems be developed that increase crash survival and prevent injuries in aircraft accidents.

References

1. SHANAHAN D.F.; SHANAHAN M. 'Kinematics of US Army helicopter crashes 1980-1985.' *Aviation Space Environ.* Med. 60 112 (1989).
2. WELLS H.V. 'The flying service from a medical point of view.' *Journal of the Royal Naval Medical Services.* 1 55 (1915).
3. WELLS H.V. Untitled. *Journal of the Royal Naval Medical Services.* 2 65 (1916).
4. ARMSTRONG H.G.; HEIM J W. 'The effect of acceleration on the living organism.' *Defence Documentation Center for Scientific and Technical Information Report ATI 25622 (1937).*
5. DE HAVEN H. 'Relationship of injuries to structure in survivable aircraft accidents.' *National Research Council Report No 440 (1945).*
6. GOERTZ, A. 'Limits and special problems in the use of ejector seats.' *Defence Research Information Centre*, DRIC-T-7684 (1954).
7. STEWART W.K. 'Ejection of pilots from aircraft – a review of the applied physiology.' *Flying Personnel Research Committee Report No 671 (1946).*
8. ANTON D.J. 'Injuries associated with escape from fast jet aircraft.' *RAF Institute of Aviation Medicine, Report No 707 (1991).*
9. STECH E.L.; PAYNE P R. 'Dynamic models of the human body.' *Aerospace Medical Research Laboratory, Wright Patterson Air Force Base, Ohio.* AAMRL-TR-66-157 (1969).
10. 'Ejection acceleration limits.' *Air Standardisation Co-Ordinating Committee (ASCC).* Air Standard 61/102/1 (1970).
11. LEWIS M.E. 'Head and spine injuries sustained on ejection from aircraft currently in service with the Royal Air Force.' *Aviat. Space Environ. Med. 71 31 (2000).*
12. BRINKLEY J.W.; SPECKER L.S.; MOSHER S.E. 'Development of acceleration exposure limits for advanced escape systems. Implications of Advanced Technologies for Air and Spacecraft Escape. *AGARD-CP-472 (1990).*
13. COGSWORTH S.C. 'Aviation pathology. Injury Prevention in Aircraft Crashes: Investigative Techniques and Applications.' *AGARD-LS-208 (1998).*
14. MCMEEKIN R.R. 'Aircraft accident investigation. In Fundamental of Aerospace Medicine.' DEHART RL (EDITOR). *Williams and Wilkins.* Baltimore, Maryland, USA (1996).