Survivable Impact Forces on Human Body Constrained by Full Body Harness

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This study was commissioned in an effort to reduce the potential for leg and ankle injury to construction and roofing workers employed in the construction and cladding of 'low roofs'. On structures of this type the workers often secure their harness lanyard to a strong point, or anchorage, at 'foot level'. If a fall should occur, the combination of 2m lanyard length plus extension of the energy absorber and the height from harness attachment to the worker's feet can exceed the height from the structural anchorage to the floor. The worker's feet may strike the ground or floor whilst the energy absorber is still deploying.

It has been suggested that reduction of the fall-arrest distance may reduce the potential for these injuries, but the laws of physics indicate this cannot be achieved without consequent increase of arrest forces on the body. This study investigates the possibility of raising the level of the fall-arrest force. It also suggests alternative solutions.

Analysis of the medical, physiological and other scientific literature regularly shows up the fact that "the learned" talk to "the learned" in terminology foreign to other educated readers. This paper seeks to 'demystify' the information and make it available and understandable to those whose interest is industrial fall-safety.

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EXECUTIVE SUMMARY

The object of this study was to determine if it is medically supportable to develop energy absorbing devices with arrest force greater than the present CEN standard 6kN maximum advised for wearers of industrial full body harnesses.

The study was initiated following reports that workers on low roofs can be exposed to a peculiar fall hazard. During steelwork and roof laying phases on low-roof constructions, e.g. factories and shops, the linear advancement of the project requires regular relocation of the worker's lanyard anchorage. In work of this type it is common for the worker to anchor to the structure at feet level. Such a low anchorage can result in ankle and leg injury if, in a fall, the combination of lanyard, energy absorber, harness assembly and body height exceeds the height to the floor or ground.

Reduction of fall height entails increase of fall arrest forces. A major feature of the study was the gathering of information on impact tolerance levels on the human body, particularly seat-to-head data.

Although there is much anecdotal information from climbing, diving, football and other sports activities, these sources were of little help due to lack of information on physical forces involved. The study therefore concentrates on relevant, scientifically measured data from biomechanical research by the National Aeronautics and Space Administration (NASA) and the Advisory Group for Aerospace and Development (AGARD) - a group serving the interests of the North Atlantic Treaty Organisation (NATO). Included in the NASA and AGARD research are medical/physiological data relating to strength of the spinal column, vertebrae and intervertebral discs. These researches were conducted largely in the 1950's and 1960's but several analyses have been produced in the subsequent years; relevant references are highlighted throughout the study.

The literature is not sympathetic to the notion of increasing present levels of arrest force on wearers of full body harnesses. A deceleration of 12G is considered survivable in a parachute harness, i.e. a harness with torso enclosing straps and shoulder straps. For such harnesses the NASA/AGARD researches indicate a 5% injury risk at 12.1G, but the differing posture, physical fitness levels, harness attachment location, 'wearer comfort' and other factors have influenced the advisability of 6G as a maximum for users of industrial harnesses.

The study includes easily understood mathematics which show that efforts to reduce arrest distance by increasing the arrest force introduce a law of 'diminishing returns'. It concludes that the present 6kN limit (EN standards) is a wise choice for body weights in the range 80kg to 100kg. But it is recommended herein that 4kN maximum arrest force is more suitable for body weights in the range 50kg to 80kg, and 8kN max would be suitable for body weights in the range 100kg to 140kg. Strong recommendations are made that UK and CEN standards bodies should seriously pursue this proposal. The initial purpose of the study - safety on low roofs - is somewhat swamped by the biomechanical information but, included in the conclusions, it is suggested that safety on low-roof work may be best improved by the use of the relatively new innovation of lightweight, portable floor mats.

Also included are guidance tables and diagrams illustrating the test performance of various systems and comparisons of these with predicted performance when used by a person. This part of the study highlights 'stretch' in various components that leads to unexpected increase in the fall-arrest height and, as such, provides guidance for installers.

In matters 'biomechanical', the work highlights the reported 'greatest risk' areas of the cervical, thoracic and lumbar vertebrae; it also identifies papers that comment on injury to internal organs at high levels of 'seat-to-head' deceleration. The reported strengths of vertebrae and intervertebral discs are shown in diagrammatic form and further listed in the annex 'Highlights of Papers'.

The study entailed the scrutiny of 53 relevant scientific and biomechanical research papers. Brief summaries of these are included as an annex, along with a full bibliography.

1 INTRODUCTION

1.1 BACKGROUND

Until the introduction of European Standards (EN standards) in 1993, the accepted UK standard for harnesses and associated equipment was BS 1397. From its inception BS 1397:1947 “Specification for safety belts and harnesses”, and its revisions over the years 1956, 1967 and 1979, advised performance norms for harnesses and associated equipment where the structural anchorage point was always above the user. The worst case was considered to be when the anchor point was horizontally in line with the attachment point of harness and lanyard (i.e. it was considered that the worker would not fall further than the length of the lanyard - 2 metres maximum – fall factor 1.0). BS 1397:1979 put a limit of 10kN on the arrest force, when tested with a 100kg articulated dummy on a fall of 2 metres.

During these development years there was awareness in the trades that a worker may not, in certain work tasks, have an anchor point above or adjacent to the harness/lanyard attachment. A fall could be from above the anchor point (e.g. a steel erector whose only strong point may, at times, be the beam beneath his/her feet). In such a case, the fall could be twice the length of the 2m long lanyard, i.e. 2 x 2m [fall factor 2.0 (FF 2.0)] = 4m.

A working group was set up by CEN (European Committee for Standardisation) in the late 1980's to rationalise the various national standards and determine fundamental norms for 'fall safety' personal protective equipment (ppe). The group investigated several issues, including the choice of test dummies (sometimes test-mass) and the consequences of 4m falls on industrial workers. The working group recommended to CEN and the EN standards committee (CEN/TC160) that a fall-arrest limit of 6kN be set for EN standards. EN 355:1992 (BS EN 355:1993) “Personal protective equipment against falls from a height – Energy absorbers” was therefore introduced, with provision for arresting a fall of 4m with a maximum arrest force of 6kN.* To accommodate the necessary extension the assembly of energy absorber and associated lanyard was permitted to extend by a maximum of 1.75m when tested with a 100kg mass on a free-fall of 4m.

Designers and manufacturers of these kinds of ppe have addressed the norms by providing energy absorbers, usually of ‘tear-stitch’ or ‘tear-ply’ construction, operating at around 4kN arrest force. At such arrest force, the energy absorbers tend to take up much of the permitted 1.75m extension. Energy absorbers designed for operation nearer the 6kN limit would be expected to require less extension in a fall-arrest event. Energy absorbers designed to operate above 6kN would require yet less extension, hence their viability is an object of this study.

> * The EN standards committee CEN/TC160 has applied the same fall-arrest force limit of 6kN for EN 353-1 and EN 353-2 arresters on rigid and flexible anchorage lines. The 6kN limit has also been applied on EN 360 retractable type fall-arresters.

1.2 OBJECTIVES

The specification for the project was “to carry out a literature search, critical review and analysis of survivable impact forces on the human body constrained by full harness”.

The work was to include an “international literature search to cover relevant medical and physiological data and opinions gathered from civilian fall-related industrial accidents, military ejection seat and parachute research (where available), environmental medicine, aviation medicine, space medicine, climbing and sailing accidents (and any other sources which may become evident in the course of the search)”.

The object of the critical review and analysis was “to accumulate information on fall arrest forces (and jolts) as these relate to injury and/or survival prospects for wearers of full body harness in accidental fall situations. Where such information or medical/physiological information is lacking, the review will address and suggest areas for further research".

Simply put, the work was intended to explore relevant international research and medical/physiological opinion and determine if these would suggest extension of the present 6kN maximum arrest force for a human subject wearing a full-body harness on a free fall of up to 4m.

1.3 DEFINITIONS

For the purposes of this report the following definitions apply, together with SI units of measurement:

1.3.1 Acceleration due to gravity (g)

Natural acceleration in a fall due to gravity. This varies slightly according to location on the planet, but is internationally accepted for calculation purposes as $9.81 m/s^2$.

1.3.2 The "G" system of units

"G" is a dimensional representation of the magnitude of acceleration, expressed as the ratio of the magnitude of measured acceleration to the magnitude of "natural" acceleration due to gravity. (A body experiences 1G [in layman's terms body weight] whilst standing immobile on the ground. This differs from the "g" system where the immobile body is described as experiencing zero "g").

1.3.3 Jolt

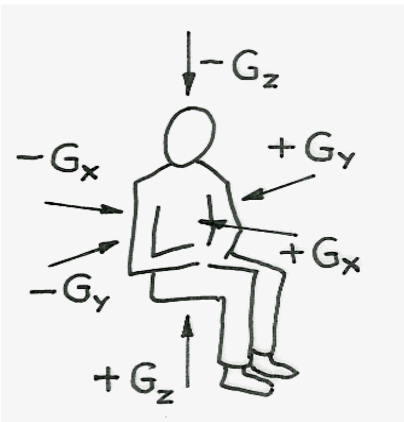
Jolt describes the rate of change, or rate of onset, of acceleration. The term is used to describe how rapidly the peak acceleration is reached. It is often expressed in the relevant biomechanical literature as G/s (G ÷ Rise Time), sometimes as m/s^3.

1.3.4 Fall Factor (FF)

This is a ratio. It is the height of a potential fall divided by the initial length of the lanyard.

1.3.5 Physiological Orientation

The system employed in this paper is the universal physiological standard system recommended by the Biodynamics Committee, AGARD Aerospace Medical Panel, *(see Snyder, "Impact"[40]).*



Upward accel. (positive G)	$+G_z$	Eyeballs down
Downward (negative G)	$-G_z$	Eyeballs up
Backward accel.	$+G_x$	Eyeballs out
Forward accel.	$-G_x$	Eyeballs in
Acceleration to right	$+G_y$	Eyeballs left
Acceleration to left	$-G_y$	Eyeballs right

Figure 1. Diagram and explanation of physiological terminology in this paper

2 A BRIEF HISTORY OF IMPACT TESTING

The major thrust in the development of impact testing came with World War II and the introduction of high-performance aircraft. Ejection seats were first installed, early in the war, in German fighter aircraft when it was found that pilots could not extract themselves manually in an emergency. It is reported that these seats were used operationally sixty times. During this time compression tests were carried out by Seigfried Ruff on cadaver vertebrae, to determine the strength of the spinal column under high positive G accelerations over short time-spans *(Henzel [22], Higgins [23])*.

> Note: References in italics and/or square brackets [-] *[-]* apply to authorities and papers summarised in Annex C "Highlights of Papers". The bibliography for this paper is to be found at Annex D.

By 1945, the Swedish air force had installed an ejection seat in their J-21 fighter. This was an aircraft with a pusher propeller and a high horizontal tailplane mounted on twin booms. With such a configuration an ejection seat was a necessity. Investigations of the strength of the spinal column were carried out by Olof Perey.

British interest in impact testing began in 1944, with studies by the Martin-Baker Aircraft Company. These were carried out with tests on an early ejection-seat test tower using dummies and human subjects. This work, at an early stage, indicated the risks of vertebral injury. Catapults were designed to lessen the risk of injury, and drogue parachutes were added to stabilise and separate the seat from the pilot. The seat was fitted with a protective visor that was pulled down by the airman at ejection and caused him to attain good spinal posture at the instant of ejection. The Martin-Baker seat was introduced into service in 1946 and its many derivatives have seen service world-wide.

The US Air Force started work on ejection seats in 1945 *(Henzel, Higgins)*. The US Navy also found it necessary to develop ejection seats for their aircraft, particularly for incidents on take-off and low-level approach to aircraft carriers. To further this work the navy developed a Rocket Assisted Propulsion Ejection Catapult (RAPEC). Researchers at Wayne State University conducted investigations in the biomechanical properties of the vertebral column, whilst teams at Massachusetts General Hospital and Massachusetts Institute of Technology worked on the properties of intervertebral discs. Several centrifuges and rocket driven sleds were built in the USA for study of tolerance and survival values for "magnitude, duration and rate of change of negative acceleration" (i.e. deceleration). Colonel J.P. Stapp is internationally known for the considerable work he carried out in biodynamics using such sleds during the 1950s.

Development of parachutes and harnesses was essential to all of the above progress in ejection seat technology. Opening of drogue and the main parachute canopy on separation of an airman from an ejection seat can cause very high deceleration forces when ejecting from a high performance aircraft. Stapp and later researchers advised 12G as an upper limit in a parachute harness [2], [6], [13], [21], [27], [40] and [43].

In time, much of the above research data found its way into the public domain via NASA and NATO scientific papers.

3 EVENTS LEADING TO CEN SELECTION OF 6kN MAXIMUM ARREST FORCE

3.1 INFORMATION GATHERING PRIOR TO EN STANDARDS

As described in the introduction, 1947 saw the beginning (in the UK) of performance testing of industrial “safety belts” and harnesses to BS1397. Until 1979 these tests were mainly “strength tests”, static and dynamic, to ensure the structural integrity of harnesses and thus to protect the wearer from falling to the ground or floor. There was no requirement to test for arrest forces.

By the 1970s there was a beginning of access to the information learned from military and aerospace studies. In the UK, the source of such information was the RAF Institute of Aviation Medicine, Farnborough *(Beeton, Ernsting, Glaister and Reader)*. The 1979 version of BS 1397 reflected this growing access in its test specification of 10G maximum for pole belts, 5G maximum for general purpose safety belts and chest harnesses, and 10G for general purpose safety harnesses. A special provision of 12.5G maximum, with full body harness, was made for coal-mine riggers who had a preference for chain lanyards. All of these 1979 version tests were carried out with an “articulated anthropometric dummy” of 100kg. ***It is noteworthy that only full body harnesses are now considered suitable in fall-risk situations.***

With the Treaty of Rome, 1975, and establishment of the European Community (later European Union) and the requirement for removal of barriers to trade, came the setting up of CEN. During the 1980's, working groups were formed with the aim of rationalising the various national standards. Their object was to facilitate the work of the later CEN technical committees and the formulation of unified standards to satisfy the European Directives. At this time the work of Stapp [43], Eiband [14] and others was becoming more accessible through NASA and NATO sources. Some of this early work was available to the CEN working group concerned with fall protection.

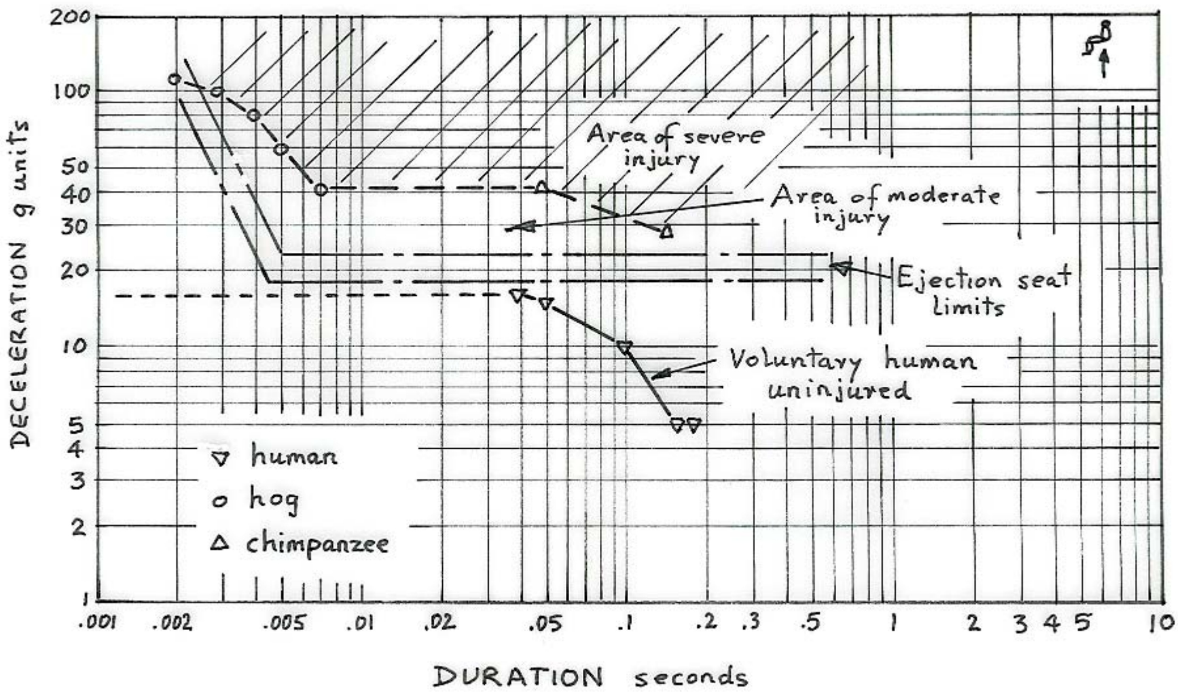


Figure 2. Survivable abrupt positive G (+G_z) impact, from Eiband

Figure 2 is a presentation of survivable abrupt positive acceleration vs. duration, based on data of Eiband. The figure uses log-log ordinate (vertical) and abscissa (horizontal) scales. Shown are the zones of voluntary human (uninjured) exposures for subjects secured with lap and shoulder straps, ejection seat design limits, and zones of probable moderate and severe injury based on research with humans, hogs and chimpanzees. Measurement of uniform deceleration of the drop test vehicle was taken at the seat level (not on the subjects).

The literature acknowledged that the uninjured, undebilitated, voluntary human exposures shown were with physically fit humans who were secured at the seat and shoulders. The available medical and physiological information related almost exclusively to studies with military personnel. Other data and medical/physiological opinion indicated that 12kN was the "maximum of tolerance" for fit men at parachute canopy opening *(Amphoux [2], Beeton [6], Delahaye [13], Hearon & Brinkley [21], Kazarian [27], Snyder [40], Stapp [43]).*

The working group (convened by Dr Maurice Amphoux) took account of the following factors:

- Military parachute harnesses are designed with greater torso constraint than industrial harnesses, i.e. there is a greater risk of upper spine (cerebral vertebrae) injury in industrial harnesses, due to flexion.

- Industrial workers, in most cases, do not have the high level of physical fitness required of military personnel.

- Industrial workers include a probably wider age-band than military personnel exposed to fall arrest risks. The literature indicated deterioration of the spine for most at ages beyond 40 years *(White & Panjabi [52], later Yoganandan [53]).*

- The proposed increase in overall fall-height from 2m to 4m would cause increase in the duration of exposure to fall-arrest forces (it should be noted that France, through AFNOR standards, had adopted 6G maximum for FF 2 with 2m long lanyards during the 1980's). Such increase in the duration of exposure to impact would project fall arrest results beyond the force/duration data of Eiband and Stapp and their definition of impact (those researchers considered 'impact' to be an event that did not exceed 0.2s).

These factors were accepted by the technical committee CEN/TC160, and 6kN was adopted as the maximum arrest force for fall-protection devices used with industrial full-body harnesses. The same norms have since been adopted for the relevant ISO standards.

3.2 U.S.A. AND CANADIAN REQUIREMENTS

To date, the US Occupational Safety and Health Authority (OSHA) [30] requirement, the American National Standard Z359 [4] and the Canadian Ontario Ministry of Labour requirement for fall protection all provide for 6ft, fall factor 1.0 falls. All three permit a maximum arrest force of 8kN.

The OSHA-required dynamic performance test for a lanyard with energy absorber employs a mass of 100kg. The free fall height is specified as "6ft (1.8m)" and the maximum permitted extension is "3.5ft (1.07m)".

The Z359 requirements are very similar. Test Mass 100kg, free fall height "6ft (1.829m)" and the maximum permitted extension "42 inches (1.067m)".

3.3 IMPLEMENTATION OF THE EN STANDARDS

Since the adoption of the EN standards, in 1993, it has been observed that the preference for arrest force - whether with energy absorbing lanyards, inertia reel (self-retracting) devices or guided type arresters - has been somewhat below the permitted 6kN level. Most UK-manufactured energy absorbers operate at around 4kN to 4.5kN, as can be seen at Annex 'A' "Force/Time traces for lanyard/energy absorber assemblies". Inertia reel devices and guided type arresters tend to operate at around 4.5kN.

These collected data on popular UK energy-absorbing lanyards are compared in figure 3 with the CEN and ISO performance norms. Also shown in the figure are the USA and Canadian norms, and the plot of the Eiband/Stapp human voluntary exposures.

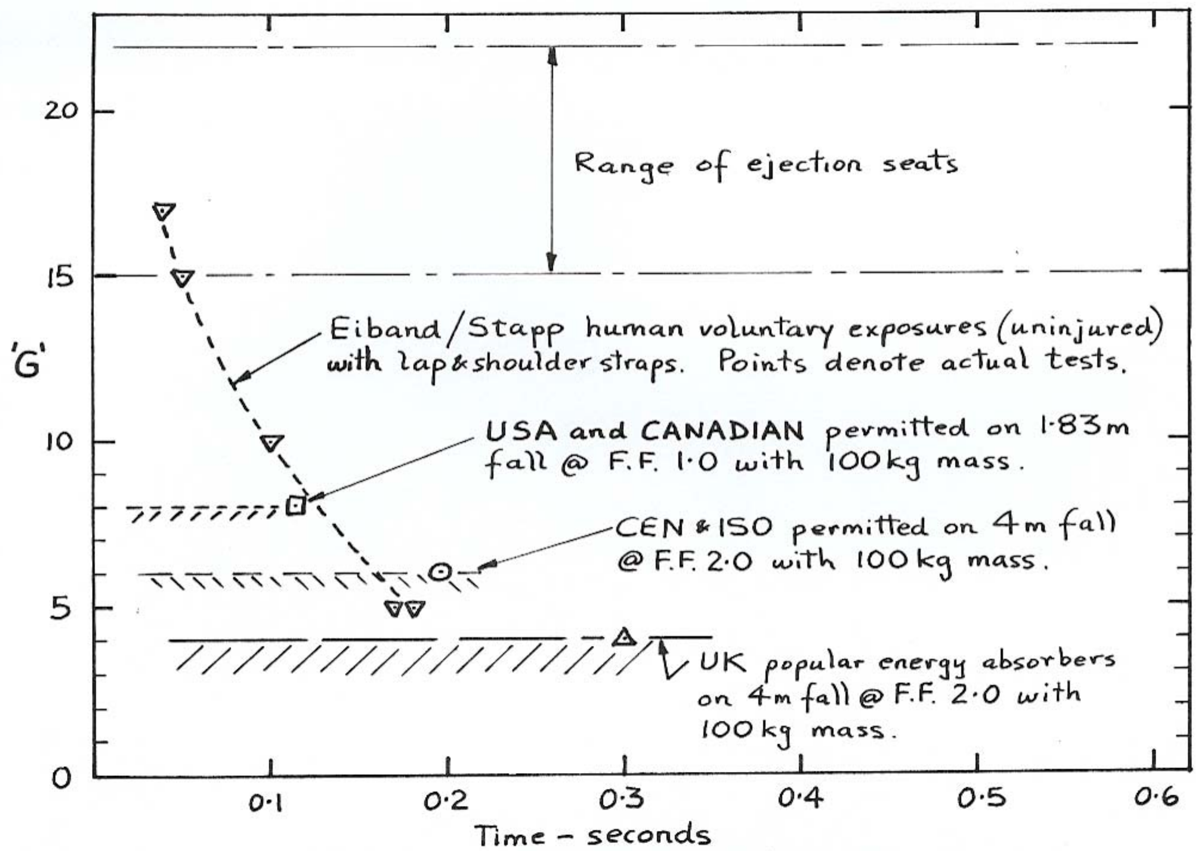


Figure 3. Comparison of fall-arrest force norms with Eiband/Stapp data

4 DATA FROM THE PRESENT STUDY

4.1 LITERATURE SEARCH

The literature search was carried out in conjunction with Knowledge Key Ltd. The databases used were:

NTIS: NATIONAL TECHNICAL INFORMATION SERVICE. Consists of summaries of US government-sponsored research, development and engineering from NASA, DOD, DOE, HUD, DOT, Dept Commerce and some 240 other agencies.

EI COMPENDEX: Electronic version of the Engineering Index, which provides abstracted information from the world's significant engineering and technological literature. The Compendex database provides world-wide coverage of approximately 4,500 journals and selected government reports and books.

SCISEARCH: Cited Reference Science Database. An international, multidisciplinary index to the Institute for Scientific Information. SciSearch contains all of the records published in the Science Citation Index plus additional records from the Current Contents publications.

TRIS: TRANSPORTATION RESEARCH INFORMATION SERVICE. Provides international coverage of ongoing research projects, published journal articles, government reports, conference proceedings and technical papers.

AEROSPACE DATABASE: Provides references, abstracts, technical documents, books, conferences and reports covering aerospace research and development in over 40 countries including Japan and Eastern European nations. It also contains reports issued by NASA, other US government agencies, international institutions, universities and private firms.

FEDERAL RESEARCH IN PROGRESS: Provides access to information about ongoing US federally funded research projects in the fields of physical sciences, engineering and life sciences.

PEDS: DEFENCE PROGRAM SUMMARIES. Collection of justification documents that correlate to each defence program element. Literally meaning Program Element Descriptive Summaries, PEDS provide detailed descriptions of specific programs and their costs.

MCGRAW-HILL COMPANIES PUBLICATIONS ONLINE: Provides text for major McGraw-Hill publications covering not only general business but also specific industries, i.e. aerospace, chemical processing, electronics and construction.

GALE GROUP AEROSPACE/DEFENCE MARKETS AND TECHNOLOGY: Provides articles and abstracts covering all aspects of the world-wide aerospace industry.

OCCUPATIONAL SAFETY AND HEALTH: Includes citations to more than 400 journal titles, as well as over 70,000 technical reports covering all aspects of health and safety.

SPORTDISCUS: The Sport Information Resource Centre, the database provider, is the largest resource centre in the world collecting and disseminating information in the area of sport and sports medicine.

MEDLINE: Is one of the major sources for biomedical literature with approximately 400,000 records added to the database per year.

INSPEC: Corresponds to the three Science Abstracts print publications: Physics Abstracts, Electrical and Electronics Abstracts, and Computer and Control Abstracts.

AGARD PAPERS: A further source of data proved to be Dstl Knowledge Services, Glasgow. This is a government office and was the source for all of the NATO AGARD scientific papers quoted and cited herein.

4.2 EFFECT OF JOLT

Jolt is defined as the rate of onset of acceleration. For the range of acceleration or deceleration considered in this study it is probably best explained as

$$\frac{\text{Initial peak acceleration}}{\text{Rise time}}$$

which can be expressed in m/s^3 or, as in most of the papers reviewed, G/s. Annex A shows several force/time traces for popular UK-manufactured lanyard/energy absorber assemblies. The jolt is seen to be the slope of the trace as the deceleration force rises on engagement of the energy absorber. The steeper the slope, the more serious the effect on the body.

The human body can tolerate very high jolt levels over very short periods of time, usually because the amplitude, or distance travelled, is small. Jolt becomes a serious problem when the duration or amplitude increases. For subjects exposed to positive G the means of constraint of the body is important.

The spinal column is limited in strength, as many of the references show *(Amphoux [2], Burton et al [7], Delahaye [12,13], Eiband [14], Higgins [23], Jones [25], Kazarian et al [27], Shaw [39], Snyder [42], Stapp [43], Swearingen et al [47], and Teyssandier [48]).* If the torso is positively secured, as for example in a modern ejection seat, the basic vertebral strength can be enhanced because the bulk of muscle and general body tissue provides support for very short periods of time *(Stapp)*. The Eiband/Stapp data indicate ejection seat operation in the range 18G to 22G. Delahaye cites "G forces greater than $+15G_z$ for duration of 0.2 to 0.5 seconds".

On separation from the ejection seat, several researchers *(Amphoux, Beeton et al [6], Delahaye, and Kazarian et al)* are of the strong opinion that 12G should be considered the maximum deceleration in a "parachute harness". It is noteworthy that US military specification 9479A considers 12.1G to be the 5% "probability of injury level*"(Kazarian et al).*

It appears that most of the workers in the area of $+G_z$ research have concentrated on tolerance of the spinal column. Where the subject individual is less securely constrained (as in an industrial harness) the risk of spinal damage increases with high G forces – and with high jolt levels! There is greater risk of flexion of the spine, and consequent injury.

There is also a risk of internal injury due to inertia of the major organs. Figure 4 illustrates how the heart and lungs are separated from the lower organs (liver, kidneys, intestines etc) by the diaphragm. In a fall-arrest event a reasonably erect spine is, to an extent, supported by the bulk of the musculo-skeletal system, but the inner organs are more 'loosely' suspended and thus more affected by jolt. Swearingen [47] tells of crashes where helicopter pilots died of ruptured aortas, and argues that 10G measured at shoulder level is a survivable maximum for helicopter pilots. Wallace and Swearingen [51] also describe a tragic case where 6 young men died in a light aircraft that hooked a power line and came to earth "in a flat attitude". None of the victims showed any external sign of injury. All had died from "severe impact trauma to internal organs (brain, heart, liver, spleen etc)". In both reports the levels of positive G and jolt were unknown, but the nature of the injuries alerts the reader to the vulnerability of the inner organs in conditions of high positive G and jolt.

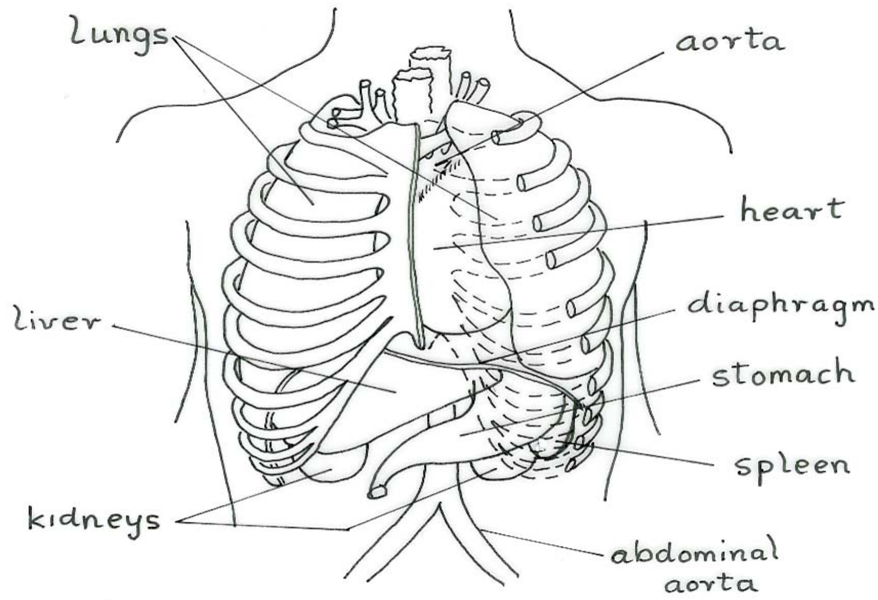


Figure 4. Major organs above and below diaphragm

4.3 STRENGTH OF VERTEBRAE AND INTERVERTEBRAL DISCS

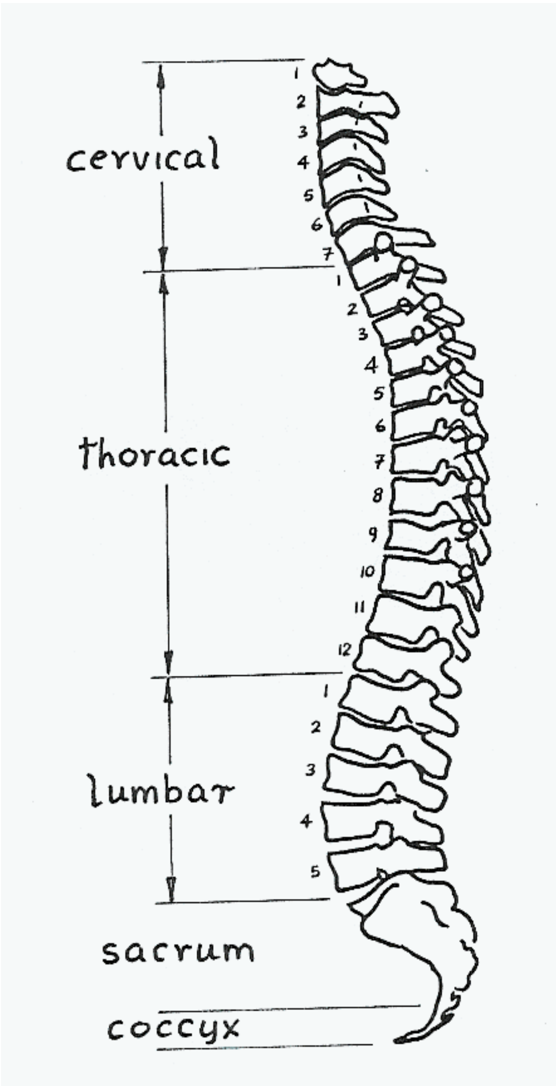


Figure 5. The Spine

Figure 5 is drawn from medical literature. It illustrates the major zones of the spinal column. The upper 7 vertebrae form the cervical spine, the next 12 the thoracic (sometimes called dorsal), and the lower 5 the lumbar. The sacrum and coccyx complete the structure. Intervertebral discs separate the vertebrae, starting at approximately 2 to 4mm thick in the cervical region and increasing to approximately 12mm thick in the lumbar region. The average length of the spinal column is of the order of 750mm, 30% to 35% composed of discs.

The strength of vertebrae increases from top to bottom of the spine, remarkably in proportion to the weight of the trunk above any given vertebra. Henzel [22] quotes Ruff and Stech findings that the resistance to positive G also rises consistently throughout the column, except for a small 'dip' in the region T12 to L2 (see also Figure 6).

The literature (see 4.4, 4.5 and 4.6) indicate C5-C7 to be most vulnerable in the upper spine, and T12-L1 to be most often damaged in the vulnerable T8-L3 lower spine range.

Figure 6 is a presentation of information on vertebral strength drawn from the literature studied. The C3 to L5 data are drawn from White and Panjabi [52], who have compared the C3 - T12 work of Messerer (1880), L1 - L5 work of Perry [in other papers Perey] (1957), and L4 work of Bell et al (1967). Higgins' [23] data from Evans et al (T12 – L5) and Massachusetts General Hospital/MIT (L2 – L5) are seen to be lower than White & Panjabi. Sances et al [36] and Teyssandier [48] are seen to arrive at higher results. Henzel [22] and Higgins separately quote data (T8 – L4) from Ruff's early work. Ruff's results from the 1940s are clearly higher than all of the other researchers. Ruff states his work to be based on body weight of 75kg. Perey identifies age difference. White and Panjabi explain the difference in results as probable differences in the design of the tests, and age and condition of cadavers. The other researchers do not appear to record age or body size. Henzel's paper is most illuminating. He compares the work of Ruff and another researcher, Stech, who noted that other serious damage occurs in vertebrae before they reach 'breaking' point. Stech had observed that vertebrae, under compression, pass through "end plate fracture", then the "limit of proportionality", next "yield point" and finally "breaking point". These features are in some ways similar to our understanding of the strength of engineering materials. End plate damage in a vertebra is caused by high disc pressures (using the analogy of a metal can, the lid and base represent the end plates).

Results on intervertebral discs have been added to illustrate the considerable strength of discs in the lower thoracic and lumbar regions. It can be seen that the discs are similar in strength to the vertebrae in the cervical and upper thoracic region. Disc strength increases progressively down the spinal column until, in the lumbar region, the discs are approximately 3 times stronger than the vertebrae *(Burton et al [7] and Sances et al [36]).* This information helps the reader see that, under impact, vertebra damage (e.g. end plate damage) can occur before disc damage (see Figure 6, Sances 'DISCS').

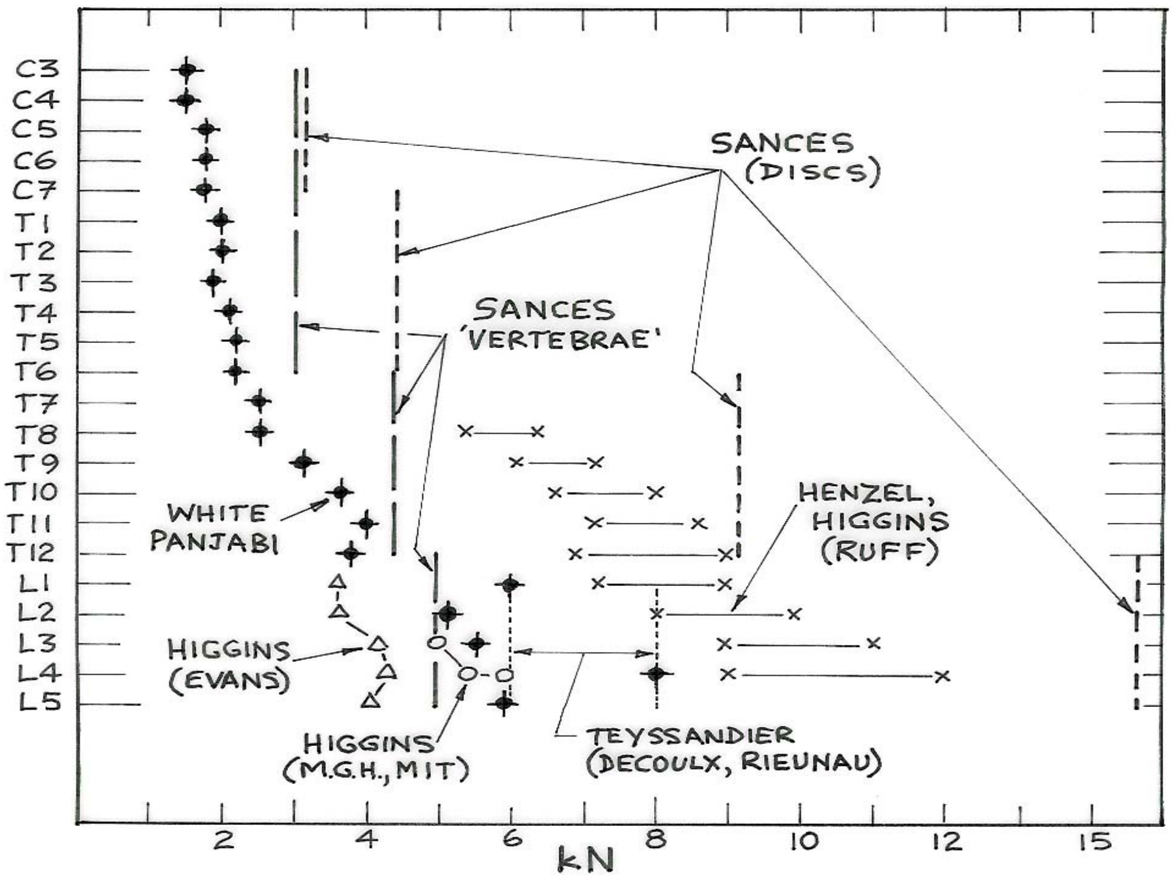


Figure 6. Strength of vertebrae and intervertebral discs

Most of the above literature is academic in nature and has been of considerable guidance in the selection of air force task tolerance levels, i.e. for physically fit personnel. Henzel argues that the setting of tolerance limits for aircraft ejection seats and subsequent parachute forces accepts that there is a 'military risk level', i.e. a risk of injury. There is no suggestion of tolerance levels for the wide variation of users of industrial harnesses.

4.4 POTENTIAL FOR CERVICAL SPINE INJURY

The cervical spine, or neck area, consists of 7 vertebrae and is the most manoeuvrable region of the spinal column. In flexion, lateral bending and torsion it is by far the most flexible. Readers who wish to pursue an academic study of the biomechanics of the cervical spine are referred to the work of Ashton-Miller & Schultz [5] and White & Panjabi [52].

Information on cervical spine injury due to impact is almost entirely drawn from military and aerospace studies. NASA has conducted considerable work in this field and much of this is to be found, in addition to European research, in papers by NATO group AGARD [Advisory Group for Aerospace Research & Development] *(AGARD paper [1] , Burton et al [7], Delahaye [13], Ewing and Thomas [16], Snyder [40], States [44]).*

The cervical spine is capable of sustaining high forces when the head and vertebrae are in optimum alignment *(Burton et al, Chub [8], Schall [37], Yoganandan [53]).* Schall describes flying manoeuvres where aircrew with helmets can be exposed to 65kgf (636N) force on the cervical vertebrae. He also notes the ability of some native Africans to carry "91kg without difficulty". When optimum alignment of the head and neck is not achieved, the potential for injury is considerable. Burton and his colleagues in NATO (RTO-TR-4) *(see Burton et al)* describe cervical spine injuries in manoeuvres of positive 4G. These usually occurred when the airmen were making observation head-movements. Deakin [11] reports similar injuries on 8G turns. It is of interest that most of the aircrew thus injured were able to return to duty after medical treatment, some to lighter duties. Burton tells of the advantages of flyers with muscular necks and, at chapter 14.7, describes effort by several air forces to introduce physical activities to strengthen neck muscles. He concludes, however, that such "an exercise regimen" has not yet been developed that will "produce balanced neck support".

Henzel [22] refers to work published by Jefferson in 1928, long before the advent of high performance aircraft (or of full-body harnesses). Jefferson had studied 2006 cases of fall-related spinal fracture. Of these, 28% involved fracture of the C5-C7 vertebrae, the lower cervical spine area (see Figure 5). This information is rather historical but, considered together with the above data, it demonstrates the vulnerability of this region of the spine to flexion. Many of the more recent researchers express concern about cases of whiplash or flexion of the upper spine, usually caused when the flyer's head and neck are not in optimum alignment at the moment of impact or high G manoeuvre. Delahaye goes further by describing hyperextension of the upper spine, suffered at times by parachutists. This is the name given to the mechanism where the inertia of the head causes the chin to strike the sternum (frontal bone of the rib cage), followed by "nodding", a to-and-fro action of the head and neck, the "classic bell-ringing" motion.

Teyssandier [49] makes the observation that cervical spine injuries among French paratroopers and sports parachutists have gradually reduced in number over the years. He associates this with developments in parachute design and folding techniques that have led to reduced "shock" at opening. His view is that the reduction in injuries is in direct ratio to the reduced "shock". Delahaye and Metges, see [13] AGARD-AG-250(Eng) Chapter 5, discuss both parachute "opening shock" and "landing shock". They observe that cervical spine injuries due to opening shock have been reduced with the introduction of "canopy first opening" instead of "canopy first packing".

In the light of these accounts, there would seem to be some concern that EN 361 "full body harnesses" permits a post-drop angle of 50^o between the "longitudinal axis of the dorsal plane of the torso dummy and the vertical". International Standard ISO 10333-1 "full-body harnesses" permits a post-fall angle of 45^o for some classes on a similar drop test. There are no data on cervical spine injury on "survived" falls with harnesses to those specifications, but such large angles from the vertical would seem to present possible whiplash and "bell-ringing" dangers. Most harness manufacturers advise adjustment of the harness so that the attachment point is high on the back, between the shoulder blades. It would, however, appear prudent to have a comprehensive investigation of the relationship between harness (plural) drop test results with torso dummies and the suspension angle for a wide range of human body size and shape. ISO 10333-1 already includes a suspension angle test with "a minimum of three people". All of the evidence points to there being a lower spinal injury rate when the spine is near vertical at the moment of impact in a harness. The ISO specification suspension procedures would appear to offer "a starting point".

4.5 POTENTIAL FOR THORACIC SPINE INJURY

The thoracic spine (T1-T12 in Figure 5), particularly T1 to T8, has a good range in torsion and flexion, but the lowest lateral bending range of the spinal column *(Ashton-Miller & Schultz [5]).*

Again, the data are virtually all from NASA and NATO military or aerospace research. Most injuries to the thoracic spine are recorded in the lower thoracic area, T8-T12, and are usually associated with injury to the lumbar L1 and L2 *(Delahaye et al [13], Henzel [22], Jones et al [25] and Snyder [41]).* Jones et al dealt with US Navy ejection injuries 1958-63 and UK aircrew ejection injuries 1949-60, all with Martin Baker seats. The ejection accelerations ranged from 17G to 22G. The paper indicates that most of the uninjured ejections were at the lower end of the $+G_z$ range, and the high proportion of injuries were at the upper end of the range. The greater proportion of injuries for US and UK aircrew were in the T8-L1 region, several flyers suffering 2 fractured vertebrae. The same study recorded Swedish injuries 1957-60 with the SAAB seat. Injuries with this seat were mainly in the T4-T7 region, with several flyers suffering 4 and 5 fractured vertebrae. There is no information on the acceleration levels for the SAAB seat, but it is to be expected that it lay within the envelope indicated at figures 2 and 3.

As with the cervical spine study, Henzel refers to Jefferson's (1928) work on the frequency of T12 and L1 fractures. Of 2006 cases, 191 (9.5%) were T12 fractures and 194 (9.7%) were L1 fractures.

The literature recommends a maximum of 12G in a parachute harness at canopy opening *(Amphoux [2], Burton et al [6], Delahaye & Metges [13], Hearon & Brinkley [21], Karazian [27], Snyder [40] and Stapp [45]).* As explained at 3.1 above, CEN accepted 6kN (i.e. 6G on a 100kg person) as a maximum for industrial workers, to take account of the wide ranges of age and physical fitness. The 6kN level of maximum arrest force, for 100kg body weight, also takes account of the probable uncontrolled nature of industrial falls and the likelihood that the spine of the subject will not be in 'optimum alignment'. But, as 5.2 herein shows, 6kN force would not appear suitable for workers whose body weight is considerably below 100kg.

4.6 POTENTIAL FOR LUMBAR SPINE INJURY

The lumbar spine (L1-L5 in Figure 5) has good flexion capacity, moderate lateral bending ability, but very low torsion capacity *(Ashton-Miller & Schultz).* Both Ashton-Miller & Schultz [5] and White & Panjabi [52] describe the considerable wall of muscle-tissue surrounding the lumbar region. These muscles are actively involved in maintaining upright stance and sitting stability. White & Panjabi also point out "they also contribute to the very high loads to which the lumbar spine is subjected". The centre of gravity of the trunk is in front of the spinal column so that considerable muscular effort is required to keep the spine in 'optimum alignment'. The muscular effort thus induces disc forces that are much higher than would be expected from the weight of trunk above the disc *(see Henzel [22] and Sances et al [36]).*

Once more, the data on lumbar spine injury and pain are virtually all from NASA and NATO military or aerospace research. The literature evidence is abundant that the intervertebral discs in this region of the spine are 3 times stronger than the vertebrae *(Burton et al [7], Sances et al).* This probably accounts for the number of researchers whose interests concentrate on the vertebral strength of the lower spine (see Figure 6). The work of all of the researchers studied *(including Henzel, Higgins, Sances, Teyssandier, White & Panjabi)* would seem to indicate that the spine, including the lumbar region, should not be capable of surviving ejection seat forces. Even the selection of 12.1G as the US military 5% "risk of injury" level in a parachute harness goes beyond the measured strength of the vertebrae. The accepted reason for the apparent strength of the lumbar spine at these accelerations is the support given by the surrounding muscular wall and other body tissue.

Shaw [39] mentions fractures of lumbar vertebrae on 25G seat-acceleration in early German work. Teyssandier [48] has conducted a survey of 1,468,399 jumps by French paratroopers. Of these, there were 219 who suffered fractures of the spine – 76% of them occurred in the lower back between T12 and L3. The reported French parachuting accidents were mainly due to 'bad landings'. Delahaye & Metges [13] discuss the effects of wind speed on parachute landing velocity. They point out that "landing shock" varies "with the square of the horizontal wind speed" and that high wind speeds account for a high proportion of landing injuries. In addition to ankle and leg injury, the spinal region at greatest risk in these circumstances is T12 to L3.

5 OBSERVATIONS ON LANYARD, ENERGY ABSORBER AND HARNESS EXTENSION IN A FALL

5.1 LANYARDS

Lanyards are generally manufactured from polyamide or polyester webbing or rope. Those made from webbing usually have very little stretch in the fall-arrest force range. Ropes are usually of 12mm or 16mm diameter and may have considerable stretch in the fall-arrest force range. Generally speaking, rope lanyards give rise to a 'gentler' increase in force on arrest. The 'stretch' time to reach peak deceleration is generally longer than for webbing, i.e. the rate of onset (or 'jolt') is lower. Where the risk analysis for a task shows there to be no danger of striking structure below the worker, the use of a rope lanyard (with energy absorber) may be preferred. If there is a need to keep 'stretch' to a minimum a webbing lanyard is probably to be preferred.

5.2 ENERGY ABSORBERS

Energy absorbers tend to be mainly of 'tear-ply' or 'tear stitch' design. Several manufacturers have produced designs that, on dynamic testing, have a reliable 'flat' peak arrest force. Test traces for three popular, different, designs are shown at Annex A. The energy absorbers and their associated webbing lanyards were dynamically tested in accordance with EN 355 (4m drop with a 100kg mass). The force/time traces have an apparently rapid rate of onset of acceleration (jolt), usual with webbing lanyards. The results were:

No.1 Peak force 4.6kN, flat 4kN arrest force, jolt 265G/s (2602m/s^3),

No.2 Peak force 4.62kN, reasonably flat 4kN arrest force, jolt 288G/s (2824m/s^3),

No.3 Peak force 4.8kN, slightly erratic arrest force between 3.5kN and 4.5kN, and jolt 118G/s (1153m/s^3).

It can be seen that the designers were "playing safe" within the 6kN limit. Nos. 1 and 2 had more rapid rates of acceleration onset (jolt) than No.3 (i.e. their 'slopes' are steeper).

Energy absorbers in Europe should comply with EN 355, which requires capability for arrest of a 100kg drop mass on a free-fall of 4m. An allowance of 1.75m is made for extension of the energy absorber in the arrest phase, thus the absorption capacity must be 5.75kJ minimum. The ISO 10333-2, Type 2, requirement is identical, whilst the Type 1 (max. free-fall 1.8m) has an arrest-force limit of 4kN and maximum extension of 1.2m (3kJ energy absorption capacity minimum).

The OSHA [30] and ANSI Z359 [4] requirements for full harness are specified as 8kN maximum with a 100kg mass free-fall of 1.8m. The maximum permitted extension is 1.07m.

The author has long been perplexed that the industry supplies only one solution for a multi-variable problem. The standards all employ a 100kg mass on dynamic testing. This does not take account of the huge range of body weight of workers in the field, nominally 50kg to 140kg (7.8st to 22st), possibly greater. The positive G effect with a 6kN energy absorber is compared with 4kN and 8kN absorbers in Table 1:

Table 1 Deceleration levels on various body weights at 6kN, 4kN and 8kN arrest force

Body mass (kg)	*6kN arrest force (G)*	*4kN arrest force (G)*	*8kN arrest force (G)*
50	12.2	8.1	16.3
60	10.2	6.8	13.6
70	8.7	5.8	11.6
80	7.6	5.1	10.2
90	6.8	4.5	9.1
100	6.1	4.1	8.1
110	5.6	3.7	7.4
120	5.1	3.4	6.8
130	4.7	3.1	6.3
140	4.4	2.9	5.8

Table 1 indicates that an energy absorber operating at 6kN maximum is probably best suited for workers of body weight 80kg to 100kg, in fall exposures of 4m (FF 2.0). In the range of body weight 50kg to 80kg the worker would be advised to seek an energy absorber of 4kN maximum. In the range of body weight 100kg to 140kg the worker would be advised to seek an energy absorber of 8kN maximum. Such a range of energy absorbers is not available to the industry at this time. (There have been 'rumblings' from the field that large people, in particular, are not well served by the present standards for energy absorbers or other fall-arresting devices designed for 'standard' dummies of 100kg). The author strongly recommends that attention be given to body weights outwith the 80kg to 100kg person. Designers should ensure that devices are capable of absorbing the potential energy of the heaviest recommended user in each range.

There is yet another factor to be considered. Few manufacturers supply information on actual test results. They may argue that there are commercial reasons for not stating the actual test performance of given energy absorbers (i.e. actual arrest force and extension), but safety engineers responsible for risk assessments would be helped considerably if this information was routinely made available.

5.3 HARNESS EXTENSION

The progression from knotted rope to waist belt to full-body harness has been a long one. Pressures of safety directives and law, along with the retirement of a reluctant older generation, have brought on the acceptance of full-body harnesses. Although designed primarily for safety, harnesses are intended to be comfortable to wear. Most are provided with ample adjustment. The growing trend in industry is to issue each worker with a 'personal' harness and other safety kit. Safety directives require that manufacturers provide full instructions on proper adjustment of a harness, but a visit to most construction sites will give the visitor an idea of how seldom the fitting instructions are heeded. The author has seen many workers with loosely fitted harness ("for comfort"), and with the lanyard attachment 'D' ring somewhere in the middle of the back, instead of between the shoulder blades per the fitting instructions.

For the best possible outcome in the event of a fall, the preferred position for attachment of lanyard to harness is atop the shoulders, as with a parachute harness. Analysis of the literature indicates that the 'upright' arrest and suspension angle will make for optimum spinal alignment, and least risk of cervical spine damage (due to whiplash or 'bell-ringing'). With this configuration the many researchers covered in the literature search agree that 12G is a "reasonable risk" maximum (the US military 5% risk of injury limit is 12.1G). It must be borne in mind that such 12G limit is for physically fit, young, military personnel.

But atop-the-shoulders attachment is not practical for users of industrial harness. It is inconvenient in most working environments and there is a general fear of injury to the head and ears in 'awkward' falls. For this reason the high dorsal (between shoulder blades) attachment position is to be preferred. In addition to being least inconvenient in carrying out a task, this attachment (on a well-designed harness that is correctly fitted) will provide reasonable spinal alignment in a fall. The same well-designed, correctly-fitted, harness will provide a satisfactory post-suspension position whilst attempting self-rescue or awaiting rescuers (but see Orzech [29] and Seddon [38] on risks when person is immobile due to injury or unconsciousness).

When adjusted correctly a harness must, of necessity, have 'slack' for body movement. Such 'slack' will usually add to the overall height of a fall (see Figure 7) when the harness straps tighten under load. This additional height from feet to harness attachment point is often missed in estimates of clearance beneath the worker. Conversely, an incorrectly fitted harness and large suspension angle may lead to a reduced height but this is not a good solution. A large angle at the trunk may cause whiplash or "bell ringing".

Pre-sternum (frontal) attachment is not addressed in this study because such attachment is normally used with vertical arrester devices with rapid 'snatch' or arrest characteristics, hence relatively short falls. In these circumstances the frontal attachment may be the most suitable solution (e.g. for ladder climbing) but a fall may expose the user to facial injury from connectors, etc..

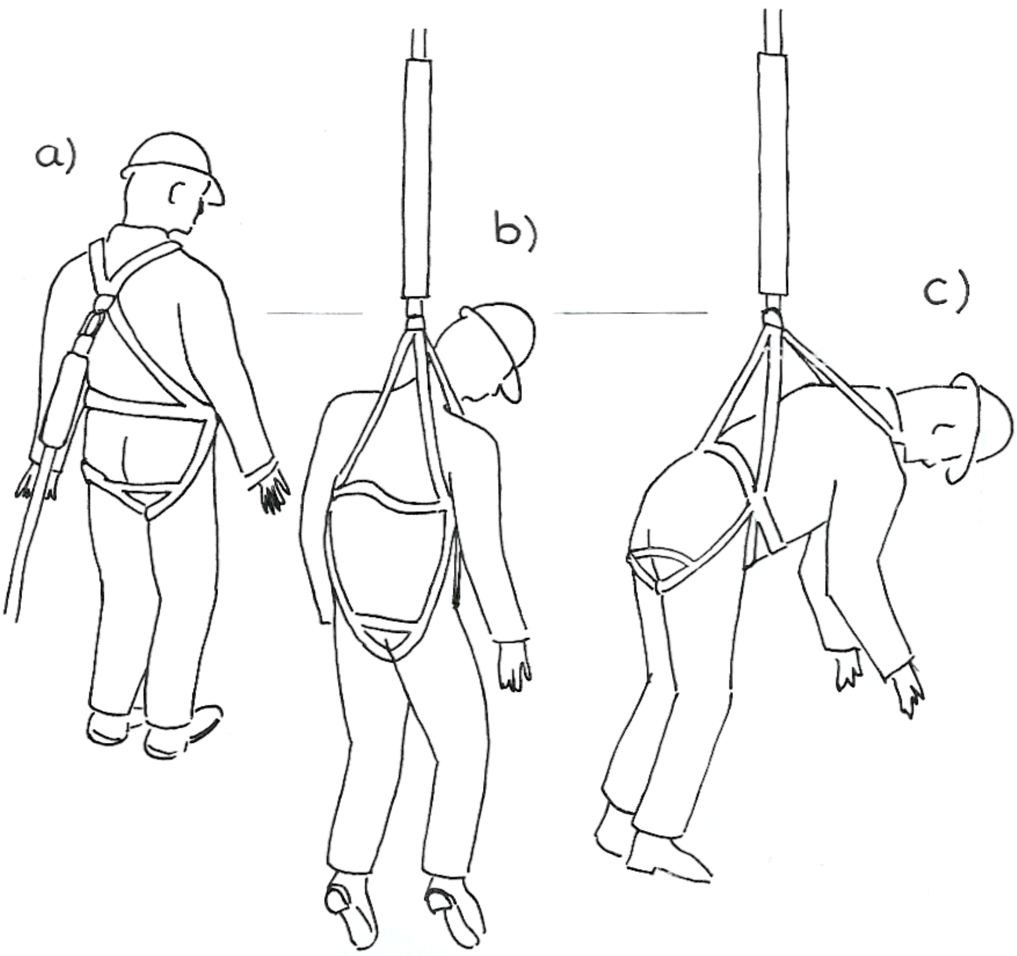


Figure 7. Illustration of harness position a) before fall, b) after fall, c) badly fitted

5.4 COMBINATION OF FACTORS AFFECTING FALL-ARREST HEIGHT

The primary objective of this study is to seek to improve safety for workers who, on 'low roofs', may have no other anchorage than the beam or structure at, or below, their feet. The height from anchorage to floor or ground level may be less than the necessary deployment length of the fall safety system. For this reason the following comments relate to the usual 'simple' safety system of connectors, lanyard, energy absorber and harness.

The factors affecting fall-arrest height are:

- Position and type of anchorage. Selection of the anchorage is determined by conditions at the work site. The anchorage may be an eyebolt or similar. It often is a wire-rope sling secured around a steelwork beam. The length of the sling is important. The author has seen slings in use that are too long for the girth of the beam. This adds to the drop height.

- Choice of material and construction for the lanyard (webbing or rope) affects the extension of the lanyard at peak arrest force. The ideal lanyard would be one that behaves 'plastically', i.e. stretches to point of peak arrest force and does not 'rebound' when the peak force is passed. It should be noted that a test house measures the extension length 'post-test', not the length at instant of peak force on a dynamic test.

- Decelerative force of the energy absorber under dynamic conditions.

- Choice of material and design of the energy absorber affects the extension of the energy absorber at peak arrest force. Energy absorbers by their very design are intended to behave as if 'plastic', but usually there is some inherent elasticity due to the material of manufacture. Again, it should be noted that a test house measures the extension length 'post-test', not the length at instant of peak force on a dynamic test.

- Selection of the harness is a company or individual choice. Proper adjustment of the harness is a very important individual skill.

- Extension of the harness at peak arrest. This is acknowledged by some in the industry, but often the amount of 'stretch' is not known. The geometry of a harness alters considerably between 'comfortable to work in' and 'suspended in' (e.g. thigh straps which are near 'horizontal' for working become 'near vertical' when the wearer is suspended in the harness (see Figure 7b). Measured from floor to harness-lanyard attachment point, many harnesses at suspension of the wearer increase that height by 200-300mm. If the attachment 'D' ring is hanging downward prior to suspension a further 100mm may be added to the height (most 'D' rings are approximately 50mm across the 'half-circle'), i.e. the apparent 'stretch' becomes 300-400mm.

- Angle of the spine at peak arrest force (not measured on test; it should be noted that the angle is measured 'post test'). The position of the lanyard attachment point on the harness is important. Figure 7c shows the effect a badly fitted harness has on the angle of suspension. Such an angle at peak force would expose the wearer to possible whiplash or 'bell-ringing' injury in a fall event.

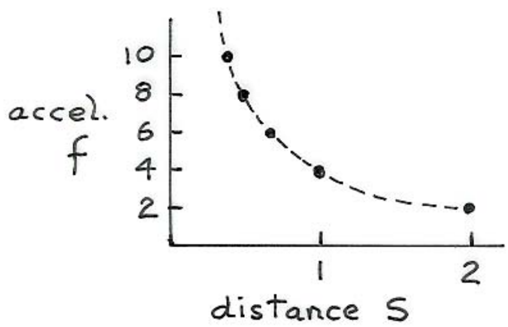
5.5 OBSERVATIONS ON FALL-ARREST CALCULATIONS

In the era of rope-only lanyards for 2m, FF 1.0 falls (e.g. BS1397:1979), it was usual to consider factors for fall-arrest forces comparing performance with torso or anthropometric dummies. This was advisable because the torso dummy behaved as a solid mass whilst the anthropometric dummy behaved as a series of hinged masses (as does the human body), the latter resulting in lower peak arrest forces. With the advent of tear-ply and similar energy absorbers with a 'flat peak' arrest force, such factors are of little consequence because the energy absorber behaves as a 'force limiter'.

The combined performance of the lanyard type, energy absorber design and harness 'stretch' is usually the major influence in the performance of the fall-arrest system. These are considered at length in Annex B and Tables 2-10 therein, but the trends can be summarised. These are:

- Deceleration distances tend to obey mathematical laws. As the deceleration force is increased, the distance travelled reduces as a geometric series. For example, in the kinematics expression

$$v^2 = 2 \times f \times s, \text{ where } v = \text{velocity}, f = \text{acceleration}, s = \text{distance}$$



as 'f' increases, 's' decreases geometrically. Thus, if we consider s = 2 when f = 2, then s = 1 when f = 4, s = 0.667 when f = 6, s = 0.5 when f = 8. The reduction in distance travelled is a 'diminishing return'. (Some deviation from this mathematical rule takes place in individual cases due to the varying performance of webbing, rope, tear-ply and harness 'stretch', hence it was deemed advisable by the author to draw up the examples shown in Annex 'B', Tables 2-10).

- Selection of a nylon rope type lanyard is a very suitable 'low jolt' solution where there is no risk of the worker hitting the structure in a fall. Its use in a 2m lanyard/energy absorber assembly may add 300-400mm to the overall fall height.

- Existing standards consider only one body weight, 100kg. An energy absorber operating at 6kN results in a 6G arrest at this body weight.

- Where a worker of body weight greater than 100kg falls on a 6kN energy absorber there is a risk that the energy absorption will be inadequate for the kinetic energy of the fall, and the worker would be exposed to arrest at much greater than 6.1G (see Annex 'B', Figure 8, page 29).

- If the body weight of the worker is substantially less than 100kg the worker will be exposed to arrest deceleration greater than 6G when using a 6kN energy absorber.

- The effect of using a 4kN energy absorber with a 100kg body weight (rather than a 6kN device) will result in a 4G peak arrest, but this can add as much as 0.5m to the overall fall-height in a FF 2.0 fall (compare Table 4 with Table 7).

- Any case for considering an 8kN energy absorber for a 4m fall (FF 2.0) of a 100kg person is weakened on learning that this would reduce the overall fall height by only approximately 0.175 m (when compared with a 6kN device – Annex 'B', compare Table 7 with Table 10).

6 CONCLUSIONS AND RECOMMENDATIONS

1. Seated, and suitably constrained, the human torso can withstand positive G 'forces' up to 40G for very short duration. The literature indicates 40G up to 50ms (0.05s). At such short duration the amplitude, or distance of arrest, is extremely small. Beyond these levels of force and duration there is considerable risk of severe injury. The literature also describes survived falls 'from great heights' into snow [Lancaster rear gunner] and paddy fields, even deliberately 'wrapped in straw bundles' and dropped into snow *(Delahaye [13], Snyder [40]),* but these are outwith the purpose of this study.

2. Ejection seats have been in use since the 1940s with ejection forces, measured at seat level, of 18G to 25G and duration up to 500ms (0.5s). Constraint during ejection is important, including head and neck. Recent designs, Martin Baker Mk16 and US models, aim at 15G to lessen risk of injury. (Note: television Channel 5 "When Pilots Eject", 27th January 2002, documented the survival story of a mid-air crash of two Russian MIG 29 fighters with model K36D ejection seats with a claimed 12G acceleration).

3. In a parachute harness, the literature supports the argument that the human torso can safely withstand fall-arrest deceleration of 12G (US military 5% risk of injury level is 12.1G). Straps must be at the shoulders. This configuration provides optimum alignment of spine.

4. The "arbitrary" selection of 6kN limit for wearers of industrial harness appears to have been a justifiable 'interim' solution in view of the preferred dorsal attachment position and its necessary deviation from 'optimum spinal alignment'. The 6kN limit applies to a body weight of 100kg, i.e. 6G maximum deceleration.

5. The medical/biomechanical literature reviewed in this study does not support an increase in arrest deceleration beyond 6G for wearers of industrial harness (though it is known that users of low body weight may be exposed to $+G_z$ arrest deceleration greater than 6G - see 7. below). In addition to the increased risk of injury, the study indicates that possible reduction in fall-arrest distance derived from increased arrest force is a 'diminishing return'.

6. It is the opinion of the author that the 6kN maximum 'one-size' energy absorber is inadequate for the wide range of body weight in industry.

7. To date, the limit of 6kN has been related to test dummies (and assumed human body weight) of 100kg. It would appear more practical to apply the 6kN limit to a range of body weight. It is suggested in this study that the range should be 80kg to 100kg. With an energy absorber operating up to 6kN, an 80kg person would experience a maximum of 7.4G whilst a 100kg person would experience a maximum of 6G (but note that such absorbers 'in fact' usually are designed to operate at 4kN to 4.5kN)

8. An energy absorber of 4kN limit would be more suitable for body weights in the range 50–80kg. At this force a 50kg person would experience 8G maximum whilst the 80kg person would experience 5G maximum. The present 6kN limit for energy absorbers could give rise to accelerations that may be injurious to persons of low body weight (a 50kg person could experience 12G).

9. For body weights in the range 100-140kg it is recommended that an energy absorber limit of 8kN would be more suitable. At this force a 100kg person would experience 7.9G maximum whilst the 140kg person would experience 5.7G maximum. The present 6kN limit for energy absorbers exposes persons of high body weight to 'run-out' of the device and consequent, possibly injurious, arrest forces when the absorber runs out of capacity (see Figure 8, Annex 'B', page 29). A pull-out force of 4kN to 4.5kN increases this risk.

10. The one-size-fits-all policy of some harness manufacturers may not be suitable for the range of body weight 50kg to 140kg. Although it may be possible for those in the wide range of body weight/size to don such a harness, the position of the harness/lanyard attachment is of paramount importance. For best performance and least risk of injury the attachment should be as high as possible between the shoulder blades. Large persons can often have difficulty attaining this 'high' attachment position in a one-size harness.

11. The one-size harness can also present difficulty for a small person. In addition to the problem of attaining the optimum attachment point there can be difficulty with fixed chest straps causing 'garrotte' effects due to harness 'stretch'.

12. The amount of 'stretch' on some designs of harness appears problematic. A harness may comply with the EN 361 test requirements but yet expose the wearer to considerable loss of 'foot clearance' in the event of a fall. There would appear to be good cause to investigate this problem further and advise standards authorities.

13. It would appear prudent to have a comprehensive investigation of the relationship between the post-drop angle on harness drop tests with torso dummies and the suspension angle for a wide range of human body size and shape in the same harnesses. To the author's knowledge large angles, e.g. 50° or 45° max. from vertical (measured post-test), were adopted in EN 361 and ISO 10333-1 respectively to admit harnesses that were at that time in wide use. The literature indicates that such large angles may expose the wearer to the possibility of whiplash or "bell ringing".

14. When the work task warrants, the engineer responsible for risk assessment should insist on obtaining 'actual' test performance data from manufacturers, particularly for lanyards and energy absorbers. The actual performance of the worker's safety system in a difficult environment is more important than a list of components with labels "complies with EN XXXX".

15. Risk assessment engineers should also be aware that rope lanyards usually 'stretch' more than do webbing lanyards. Such increased stretch gives rise to reduced 'jolt' at onset of arrest, but the total extension length should be taken into account where there is risk of the worker striking parts of the structure below the anchor point.

16. The above suggestions may improve safety in work situations where 'foot clearance' and safety-system performance can be known or estimated. Such knowledge may not be available on many 'low roof' applications. Other means of personnel protection may be necessary.

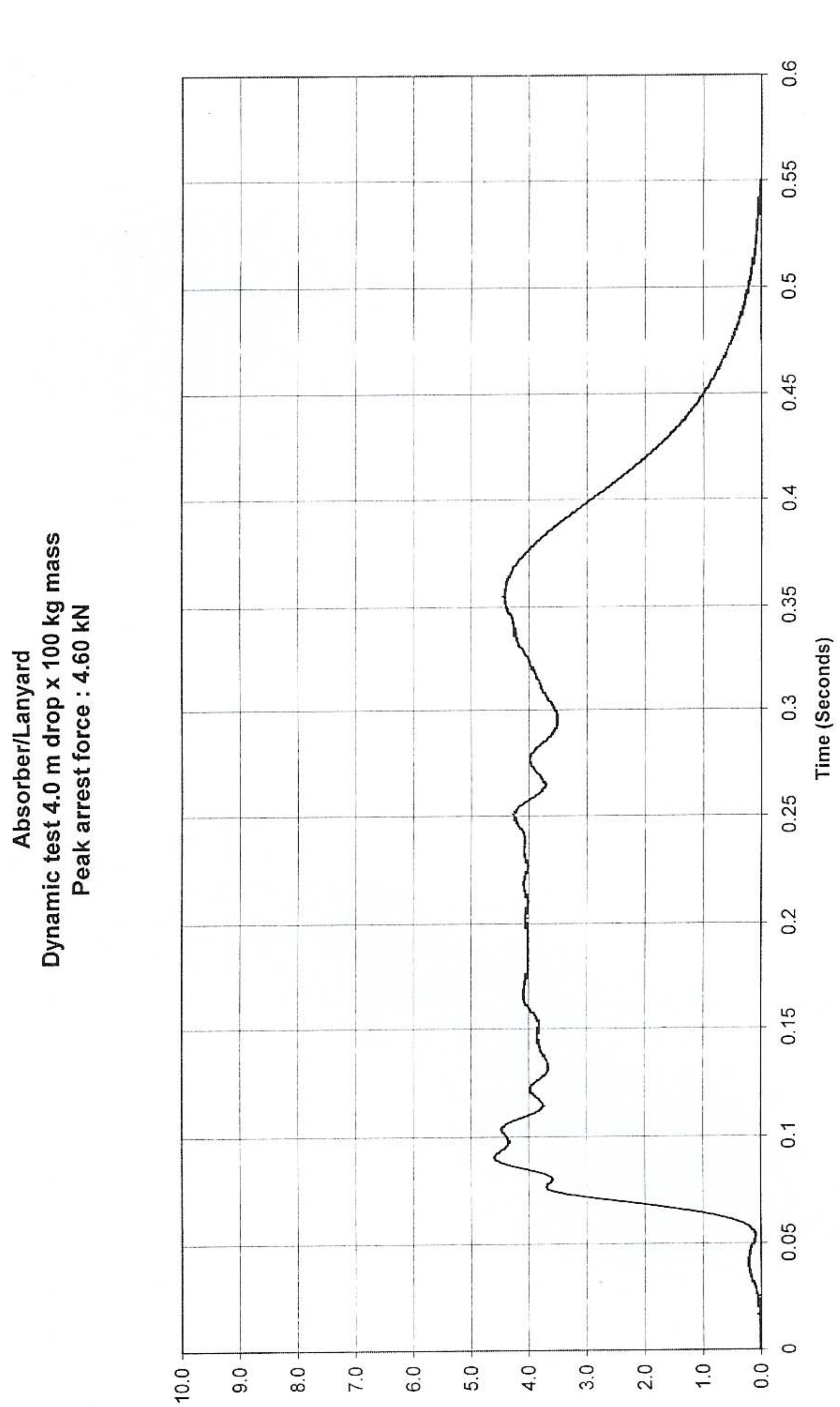
17. In some applications it may be necessary to consider nets.

18. In some applications it may be advisable to consider floor mats. Floor mats are now used by some housing developers to safeguard workers who may be in danger of falling through uncovered joists or roof trusses to the floor level below. The mats are portable and lightweight and can be moved from floor-to-floor.

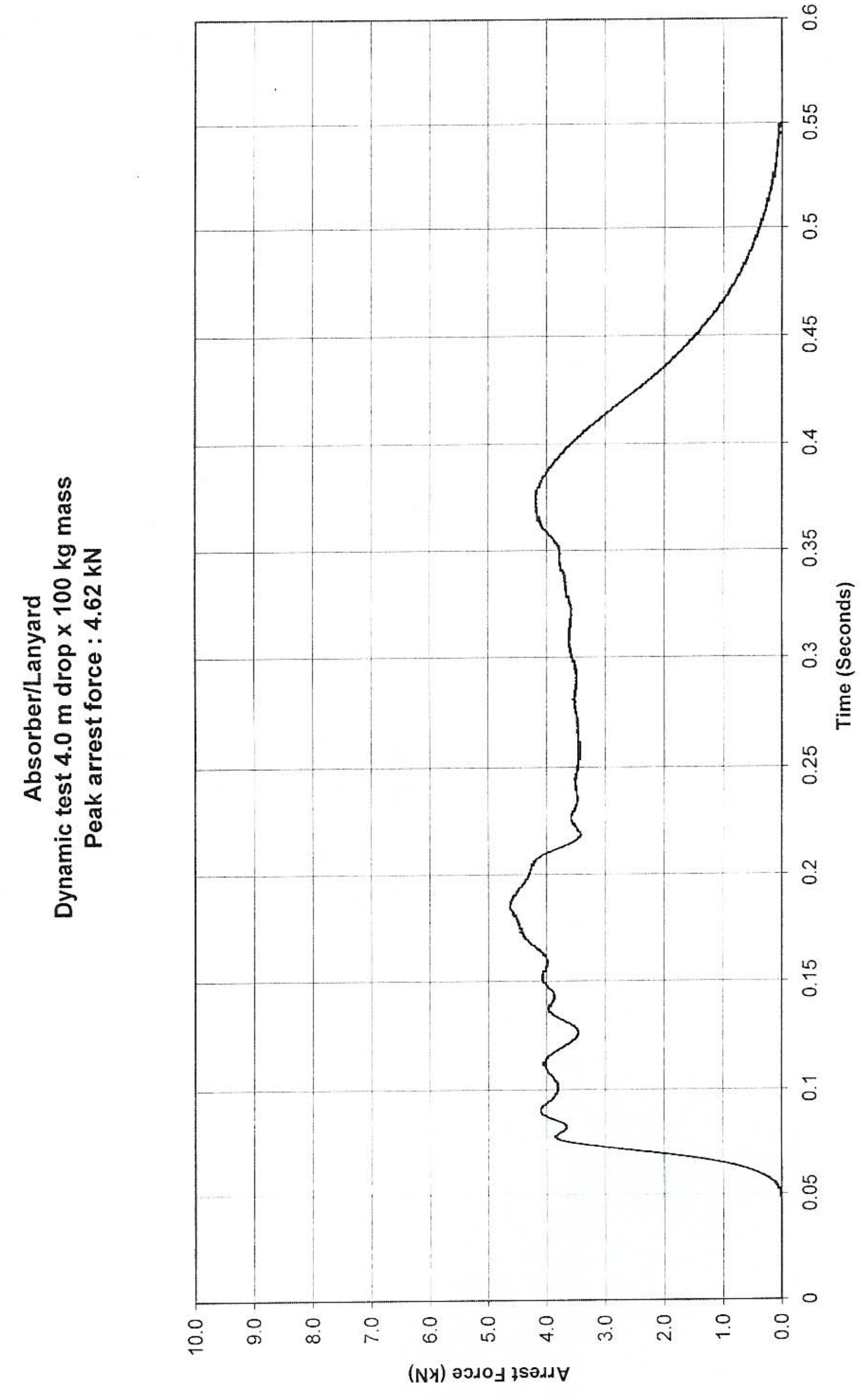
7 ANNEX 'A'

FORCE/TIME TRACES FOR LANYARD/ENERGY ABSORBER ASSEMBLIES

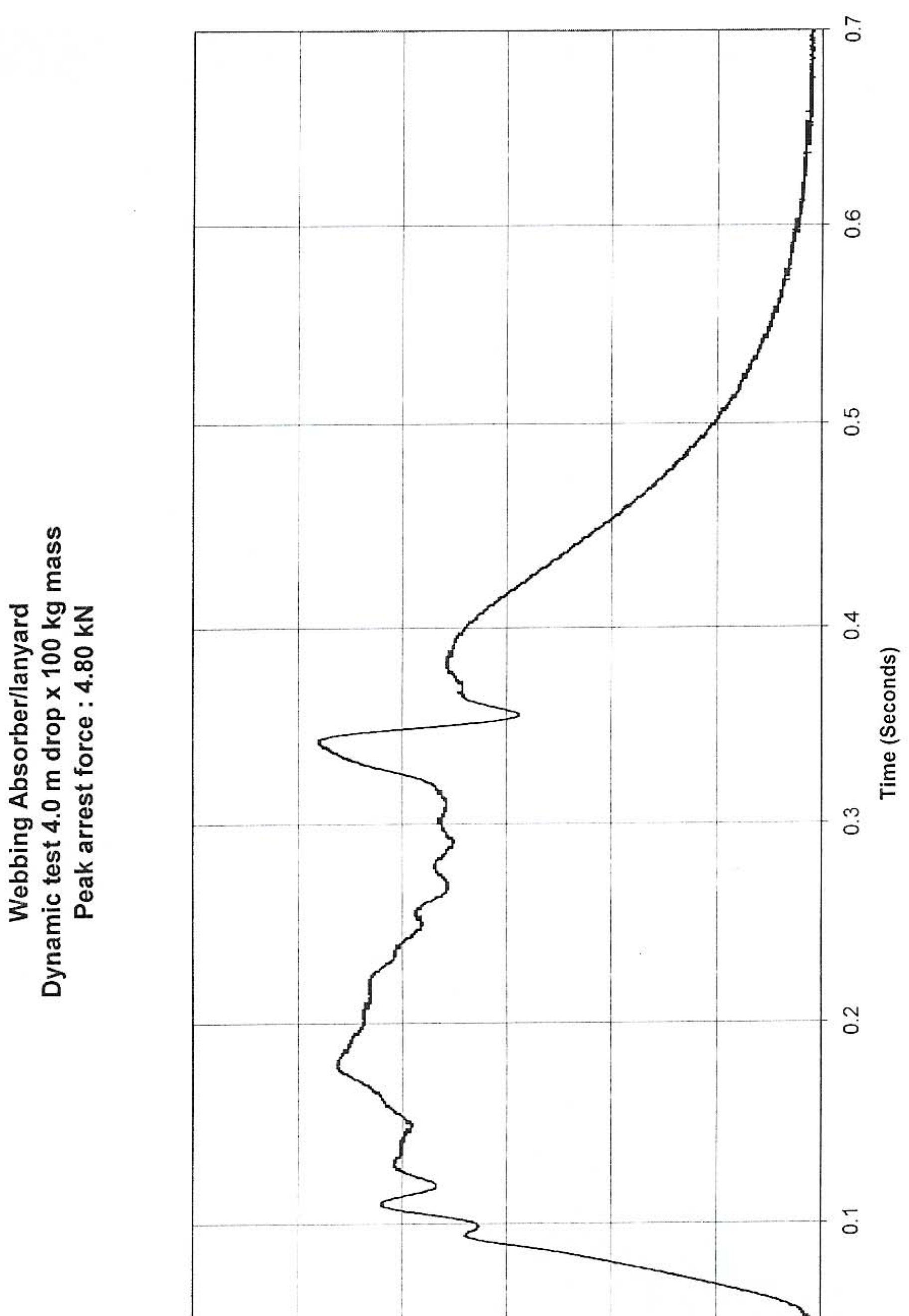
This Annex reprints the force/time traces for three popular, UK-manufactured, combined lanyard and energy absorber assemblies. The identities of the manufacturers have been withheld.


Absorber/Lanyard
Dynamic test 4.0 m drop x 100 kg mass
Peak arrest force : 4.60 kN
Arrest Force (kN)
Time (Seconds)

Trace No 1


Absorber/Lanyard
Dynamic test 4.0 m drop x 100 kg mass
Peak arrest force : 4.62 kN
Arrest Force (kN)
Time (Seconds)
0.0
1.0
2.0
3.0
4.0
5.0
6.0
7.0
8.0
9.0
10.0
0
0.05
0.1
0.15
0.2
0.25
0.3
0.35
0.4
0.45
0.5
0.55
0.6

Trace No 2


Webbing Absorber/lanyard
Dynamic test 4.0 m drop x 100 kg mass
Peak arrest force : 4.80 kN
Arrest Force (kN)
6.0
5.0
4.0
3.0
2.0
1.0
0.0
0
0.1
0.2
0.3
0.4
0.5
0.6
0.7
Time (Seconds)

Trace No 3

8 ANNEX B

TABLES OF CALCULATED DROP HEIGHTS FOR VARIOUS LANYARD AND ENERGY ABSORBER COMBINATIONS

This annex resulted from a desire to help the reader understand the potential difference between test house results and the probable performance of simple systems in a fall event. EN 355 tests on energy absorbers are carried out with a deadweight. In some cases the test may include chain to make up the required 2m test length (i.e. there may be no guidance on webbing or rope lanyard 'stretch' in the test report).

The tables herein estimate total fall height using a combination of webbing lanyard plus energy absorber and deadweight, and compare this with the fall height using a combination of 12mm nylon, 3 strand, rope lanyard plus energy absorber and deadweight. This was expanded to include webbing lanyard plus energy absorber and human subject with harness, compared with 12mm nylon rope lanyard plus energy absorber and human subject with harness. For simplicity the same webbing lanyard properties are assumed throughout, viz. 30kN breaking strength, 15% extension to failure (typical test result). Similarly, the properties of the 12mm nylon rope are assumed throughout, viz. 30kN breaking strength and 45% extension to failure (typical test result). Throughout the calculations for these tables the deadweight is assumed to be 100kg and the human subject 100kg weight. The tables should not be taken as 'guides' for all fall arrest situations. They are estimates of fall heights in the stated circumstances and are provided to enable the reader to understand the differences that may arise between the results of 'standards' testing and actual use by human subjects.

Tables 2-4 consider fall-arrests with an energy absorber of 4kN 'flat' performance on falls of 2m (FF1.0), 3m (FF1.5) and 4m (FF2.0) respectively. The structural anchorage is assumed to be rigid in every case.

Looking to Table 2, first column, the measurements taken by the test house prior to the drop test would be $L_1 + E_1 + D = H_{d1}$ which would be compared with the post-test measurement, H_e. In fact, the assembly would have extended at peak force to $L_2 + E_2 + D = H_{d2}$ due to elasticity in the system. Thus the maximum extension is expressed as (webbing lanyard + energy absorber + deadweight) 3.322m minus 2.60m, i.e. extension at peak force = 0.722m (compared with H_e minus 2.60, i.e. 0.688m measured post-test).

It is thus seen that the measurement taken by the test house (per the test specification) when the deadweight is suspended post-test is, strictly speaking, not an accurate statement of the maximum extension of the assembly. Measurement at maximum elasticity of the assembly is expensive, requiring high-speed measurement and sophisticated recording equipment, and is not routinely carried out by test houses.

In Table 2, second column (Ø12 nylon rope lanyard + energy absorber + deadweight), the maximum extension is 1.085m, illustrating the fact that the nylon rope is more elastic than webbing (note that measured post-test the extension would appear to be 3.54 - 2.60, i.e. 0.94m).

In Table 2, third column (webbing lanyard + energy absorber + harness + human subject), the maximum extension is seen to be 1.238m measured at the feet (1.168m post-test). This may seem large when compared with the first column but it should be noted that extension of the harness and the resultant increase in kinetic energy have been taken into account.

Table 2, fourth column (Ø12 nylon rope lanyard + energy absorber + harness + human), shows an extension of 1.599m measured at the feet (1.429m post-test).

Summarising Table 2 (2m free fall with 4kN energy absorber),

Extension (webbing lanyard + energy absorber + deadweight) = 0.722m
" (" " + " " + harness + human) = 1.238m
" (Ø12 nylon rope lanyard + energy absorber + deadweight) = 1.085m
" (" " " " + " " + harness + human) = 1.599m

The difference in extension from deadweight to human subject is due to 'stretch' of the harness and the effect of the additional fall height/kinetic energy on the total system.

Summarising Table 3 (3m free fall with 4kN energy absorber),

Extension (webbing lanyard + energy absorber + deadweight) = 1.057m
" (" " + " " + harness + human) = 1.568m
" (Ø12 nylon rope lanyard + energy absorber + deadweight) = 1.419m
" (" " " " + " " + harness + human) = 1.929m *

* The post-test extension 'at the feet' with this system would be of the order of 1.759m, exceeding the 1.75m limit for the test with the 100kg deadweight.

Summarising Table 4 (4m free fall with 4kN energy absorber),

Extension (webbing lanyard + energy absorber + deadweight) = 1.391m
" (" " + " " + harness + human) = 1.90m *
" (Ø12 nylon rope lanyard + energy absorber + deadweight) = 1.752m **
" (" " " " + " " + harness + human) = 2.261m #

* The post-test dimension would be of the order of 1.83m, exceeding the 1.75m limit.
** The post-test dimension would be of the order of 1.607m.
# The post-test result for Ø12 nylon rope lanyard + energy absorber + harness + human would be of the order of 2.091m, exceeding the 1.75m limit.

Tables 5 to 7 deal with 2m, 3m and 4m falls on a 6kN energy absorber.

Summarising Table 5 (2m free fall with 6kN energy absorber),

Extension (webbing lanyard + energy absorber + deadweight) = 0.462m
" (" " + " " + harness + human) = 0.954m
" (Ø12 nylon rope lanyard + energy absorber + deadweight) = 0.903m
" (" " " " + " " + harness + human) = 1.396m

The difference in extension from deadweight to human subject is due to 'stretch' of the harness and the effect of the additional fall height/kinetic energy on the total system.

Summarising Table 6 (3m free fall with 6kN energy absorber),

Extension (webbing lanyard + energy absorber + deadweight) = 0.662m
" (" " + " " + harness + human) = 1.154m
" (Ø12 nylon rope lanyard + energy absorber + deadweight) = 1.103m
" (" " " " + " " + harness + human) = 1.595m

Summarising Table 7 (4m free fall with 6kN energy absorber),

Extension (webbing lanyard + energy absorber + deadweight)	= 0.862m
" (" " + " " + harness + human)	= 1.354m
" (Ø12 nylon rope lanyard + energy absorber + deadweight)	= 1.303m
" (" " " " + " " + harness + human)	= 1.795m *

* But the post-test extension for Ø12 nylon rope lanyard + energy absorber + harness + human would be of the order of 1.575m.

Comparison of the fourth column of Table 7 with the fourth column of Table 4 (i.e. Ø12 nylon rope lanyard + energy absorber + harness + human subject) indicates a reduction in extension of 2.261m – 1.795m = 0.466m in going from 4kN arrest to 6kN arrest force.

Tables 8 to 10 deal with 2m, 3m and 4m falls on an 8kN energy absorber

Summarising Table 8 (2m free fall with 8kN energy absorber),

Extension (webbing lanyard + energy absorber + deadweight)	= 0.354m
" (" " + " " + harness + human)	= 0.845m
" (Ø12 nylon rope lanyard + energy absorber + deadweight)	= 0.845m
" (" " " " + " " + harness + human)	= 1.336m

The difference in extension from deadweight to human subject is due to 'stretch' of the harness and the effect of the additional fall height/kinetic energy on the total system.

Summarising Table 9 (3m free fall with 8kN energy absorber),

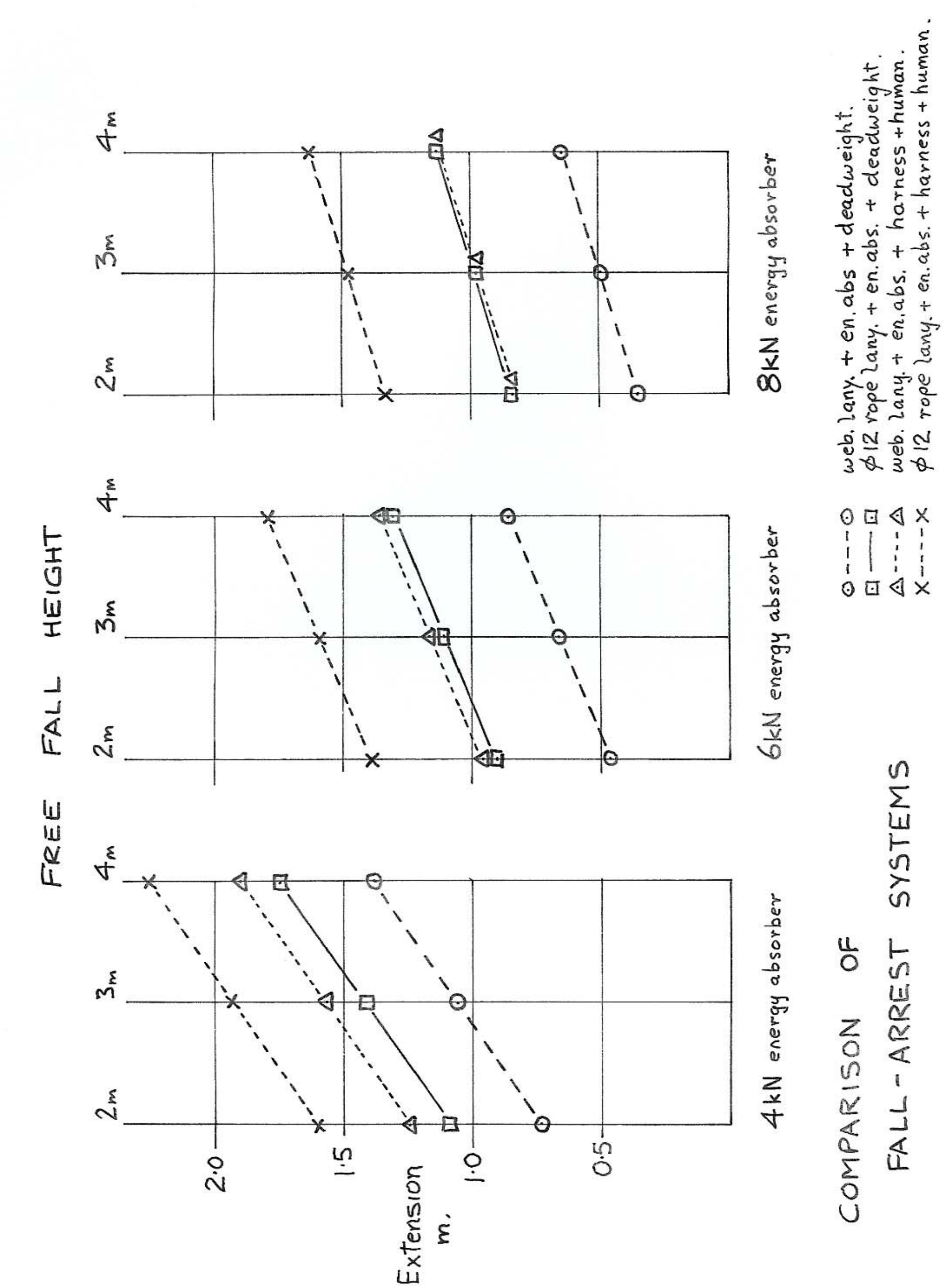
Extension (webbing lanyard + energy absorber + deadweight)	= 0.496m
" (" " + " " + harness + human)	= 0.988m
" (Ø12 nylon rope lanyard + energy absorber + deadweight)	= 0.987m
" (" " " " + " " + harness + human)	= 1.479m

Summarising Table 10 (4m free fall with 8kN energy absorber),

Extension (webbing lanyard + energy absorber + deadweight)	= 0.639m
" (" " + " " + harness + human)	= 1.130m
" (Ø12 nylon rope lanyard + energy absorber + deadweight)	= 1.130m
" (" " " " + " " + harness + human)	= 1.621m

Comparison of the fourth column of Table 10 with the fourth column of Table 7 (i.e. Ø12 nylon rope lanyard + energy absorber + harness + human subject) indicates a reduction in extension of 1.795m – 1.621m = 0.174m in going from 6kN arrest to 8kN arrest force.

These comparisons are shown in diagrammatic form below:



The calculations therefore indicate a substantial reduction in assembly extension in going from 4kN arrest force to 6kN arrest force, 0.466m reduction. The same is not true when going from 6kN to 8kN. The reduction in this case is only a further 0.174mm. This argument alone would appear to weaken the case for increasing arrest force in order to reduce the arrest distance. The gain is not sufficient to justify the change (diminishing return).

Of much greater importance is the case for designing energy absorbers to match body weight, as argued at 5.2. There it is stated "that an energy absorber operating at 6kN maximum is probably best suited for workers of body weight 80kg to 100kg in fall exposures of 4m (FF 2.0). In the range of body weight 50kg to 80kg the worker would be advised to seek an energy absorber of 4kN maximum". Workers of body weight 100kg to 140kg would be advised to seek an energy absorber of 8kN maximum. In designing an energy absorber it will be necessary to ensure that the device will absorb the kinetic energy of the largest recommended user without risk of a "bump" at full deployment of the tear-ply or other mechanism. There has been a risk with some energy absorbers (designed to comply with the 6kN 'maximum') that their operation at 4kN, though apparently 'soft' for a large worker, has led to total deployment of the absorber and transition to a 'lanyard only' arrest, with consequent increase in arrest force. Figure 8 illustrates a case - an 'apparently soft' energy absorber has insufficient energy absorption capacity at 4kN. The excess kinetic energy is absorbed by the webbing lanyard, with consequent increase of arrest force.

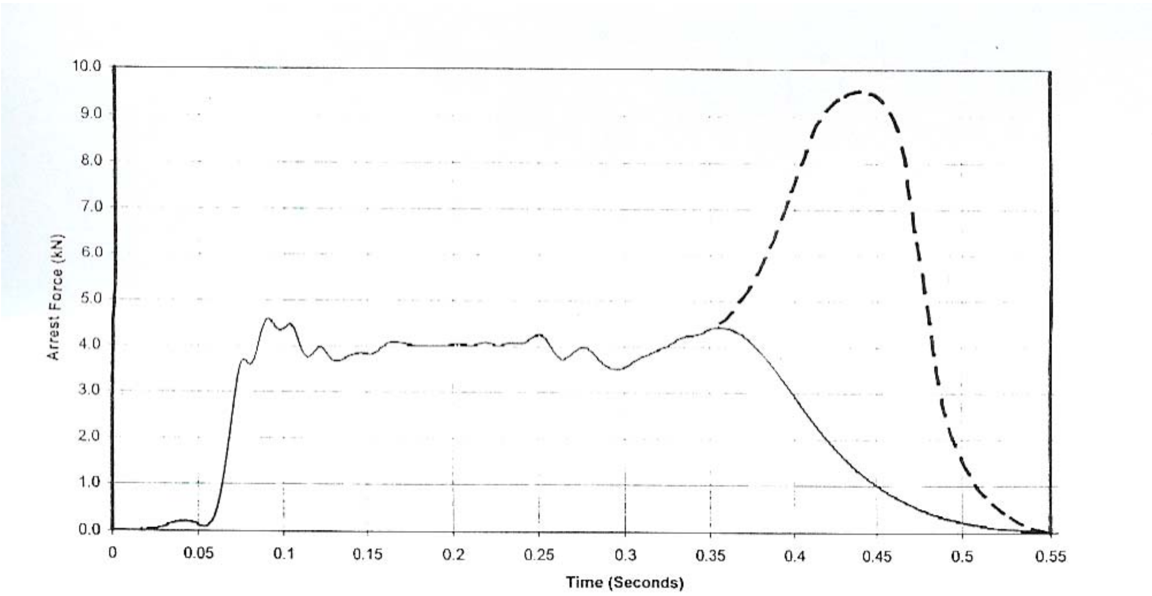
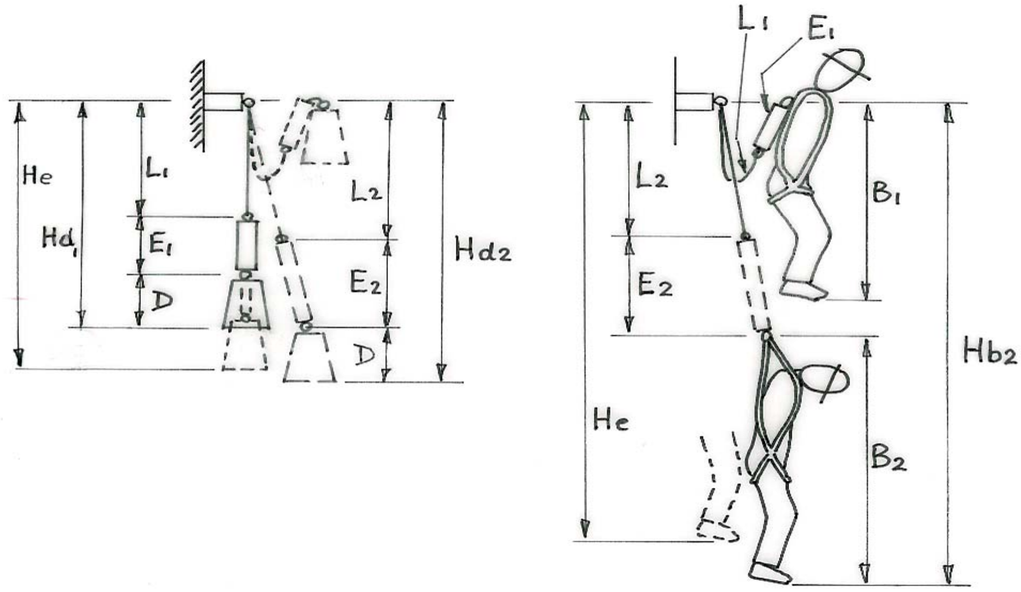


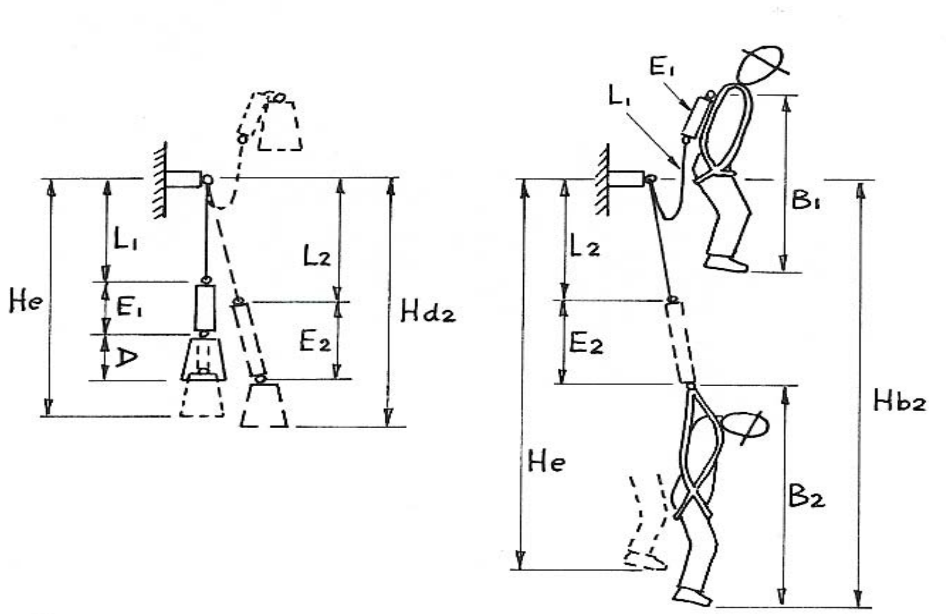
Figure 8 Illustration of increase in arrest force when the falling person has fully deployed an energy absorber



100kg Deadweight, 100kg Human,
2m Free Fall (FF 1.0), 4kN Energy Absorber

Table 2 (ref. Annex B)

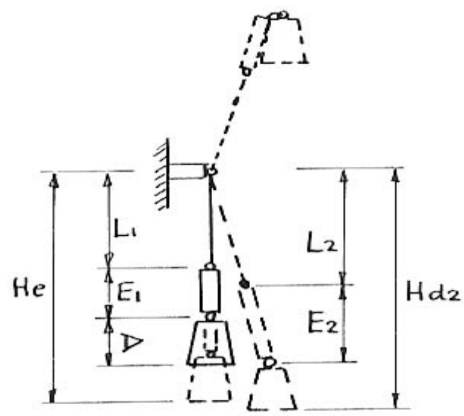
TABLE 2	Webbing Lanyard and Deadweight	Ø12 Nylon Rope Lanyard and Deadweight	Webbing Lanyard and Human	Ø12 Nylon Rope Lanyard and Human
Lanyard Length Including Karabiners L_1	1.65	1.65	1.65	1.65
Energy Absorber Initial Length E_1	0.35	0.35	0.35	0.35
Height of Deadweight D	0.60	0.60	-	-
Height – Boots to Harness Attachment B_1	-	-	1.50	1.50
$L_1 + E_1 + D = H_{d1}$ (initial length)	2.60	2.60	-	-
$L_1 + E_1 + B_1 = H_{b1}$ (initial)	-	-	3.50	3.50
Lanyard Length at Max Arrest L_2	1.694	1.965	1.694	1.965
Energy Absorber Length At Max Arrest E_2	1.028	1.120	1.159	1.249
Height – Boots to Harness Attachment at Max Arrest B_2	-	-	1.885	1.885
$L_2 + E_2 + D = H_{d2}$ (worst case)	3.322	3.685	-	-
$L_2 + E_2 + B_2 = H_{b2}$ (worst case)	-	-	4.738	5.099
Extension $H_{d2} – H_{d1}$	0.722	1.085	-	-
Extension $H_{b2} – H_{b1}$	-	-	1.238	1.599
At Equilibrium H_e	3.288	3.540	4.668	4.929
Duration at 4kN	186ms	198ms	203ms	214ms

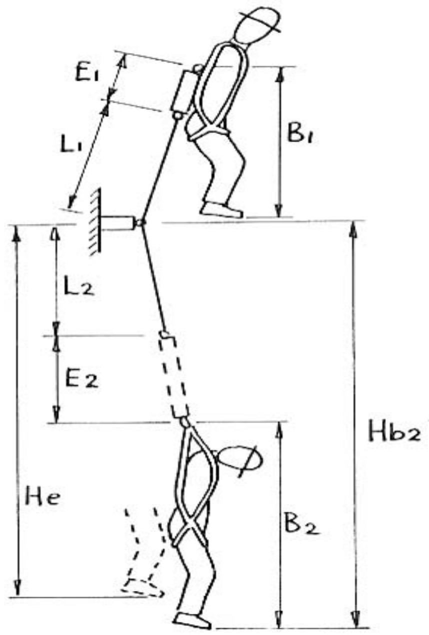


100kg Deadweight, 100kg Human,
3m Free Fall (FF 1.5), 4kN Energy Absorber

Table 3 (ref. Annex B)

TABLE 3	Webbing Lanyard and Deadweight	Ø12 Nylon Rope Lanyard and Deadweight	Webbing Lanyard and Human	Ø12 Nylon Rope Lanyard and Human
Lanyard Length Including Karabiners L_1	1.65	1.65	1.65	1.65
Energy Absorber Initial Length E_1	0.35	0.35	0.35	0.35
Height of Deadweight D	0.60	0.60	-	-
Height – Boots to Harness Attachment B_1	-	-	1.50	1.50
$L_1 + E_1 + D = H_{d1}$ (initial length)	2.60	2.60	-	-
$L_1 + E_1 + B_1 = H_{b1}$ (initial)	-	-	3.50	3.50
Lanyard Length at Max Arrest L_2	1.694	1.965	1.694	1.965
Energy Absorber Length At Max Arrest E_2	1.363	1.454	1.489	1.579
Height – Boots to Harness Attachment at Max Arrest B_2	-	-	1.885	1.885
$L_2 + E_2 + D = H_{d2}$ (worst case)	3.657	4.019	-	-
$L_2 + E_2 + B_2 = H_{b2}$ (worst case)	-	-	5.068	5.429
Extension $H_{d2} - H_{d1}$	1.057	1.419	-	-
Extension $H_{b2} - H_{b1}$	-	-	1.568	1.929
At Equilibrium H_e	3.623	3.874	4.998	5.259
Duration at 4kN	227ms	237ms	241ms	250ms

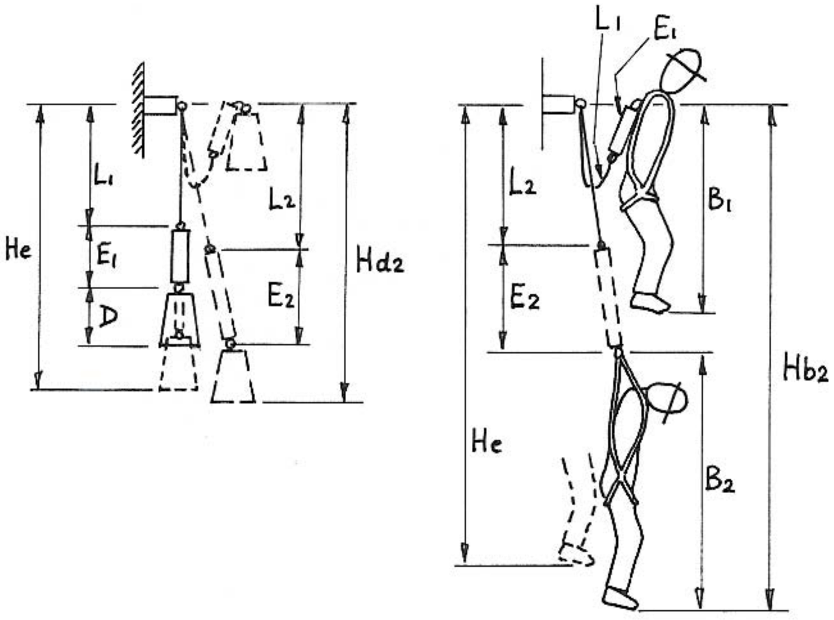




100kg Deadweight, 100kg Human,
4m Free Fall (FF 2.0), 4kN Energy Absorber

Table 4 (ref. Annex B)

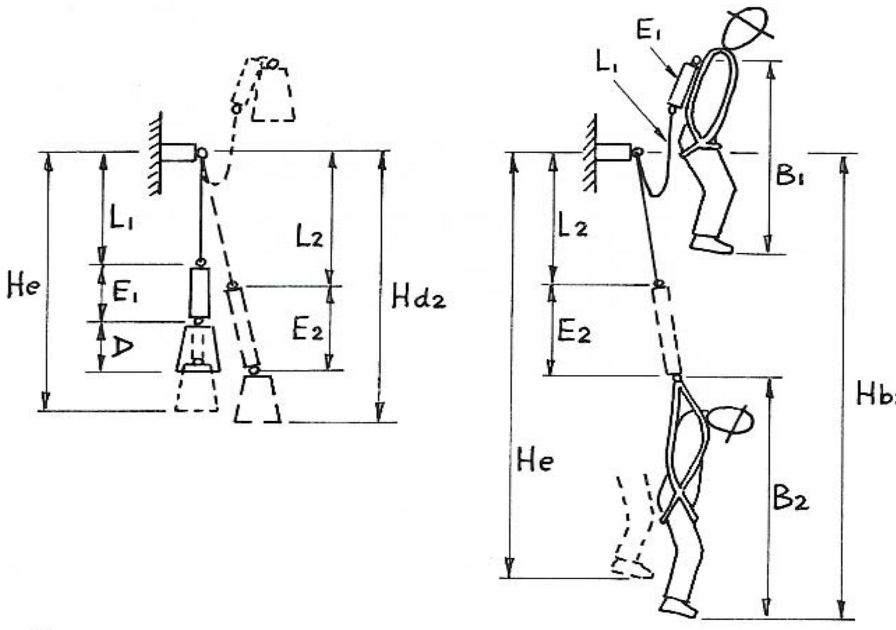
TABLE 4	Webbing Lanyard and Deadweight	Ø12 Nylon Rope Lanyard and Deadweight	Webbing Lanyard and Human	Ø12 Nylon Rope Lanyard and Human
Lanyard Length Including Karabiners L_1	1.65	1.65	1.65	1.65
Energy Absorber Initial Length E_1	0.35	0.35	0.35	0.35
Height of Deadweight D	0.60	0.60	-	-
Height – Boots to Harness Attachment B_1	-	-	1.50	1.50
$L_1 + E_1 + D = H_{d1}$ (initial length)	2.60	2.60	-	-
$L_1 + E_1 + B_1 = H_{b1}$ (initial)	-	-	3.50	3.50
Lanyard Length at Max Arrest L_2	1.694	1.965	1.694	1.965
Energy Absorber Length At Max Arrest E_2	1.697	1.787	1.821	1.911
Height – Boots to Harness Attachment at Max Arrest B_2	-	-	1.885	1.885
$L_2 + E_2 + D = H_{d2}$ (worst case)	3.991	4.352	-	-
$L_2 + E_2 + B_2 = H_{b2}$ (worst case)	-	-	5.400	5.761
Extension $H_{d2} – H_{d1}$	1.391	1.752	-	-
Extension $H_{b2} – H_{b1}$	-	-	1.900	2.261
At Equilibrium H_e	3.957	4.207	5.330	5.591
Duration at 4kN	262ms	270ms	274ms	282ms



100kg Deadweight, 100kg Human,
2m Free Fall (FF 1.0), 6kN Energy Absorber

Table 5 (ref Annex B)

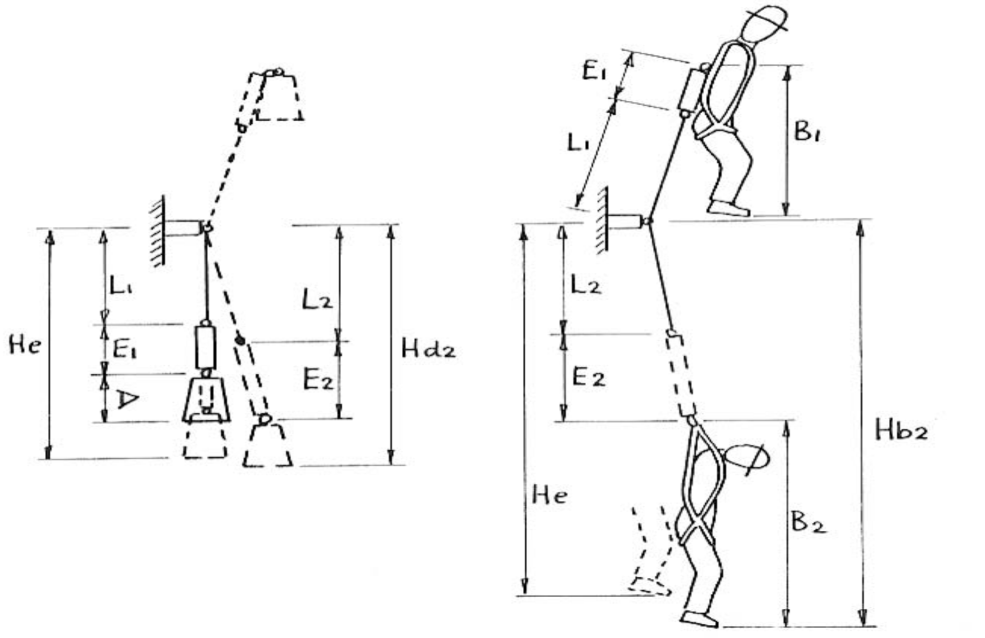
TABLE 5	Webbing Lanyard and Deadweight	Ø12 Nylon Rope Lanyard and Deadweight	Webbing Lanyard and Human	Ø12 Nylon Rope Lanyard and Human
Lanyard Length Including Karabiners L_1	1.65	1.65	1.65	1.65
Energy Absorber Initial Length E_1	0.35	0.35	0.35	0.35
Height of Deadweight D	0.60	0.60	-	-
Height – Boots to Harness Attachment B_1	-	-	1.50	1.50
$L_1 + E_1 + D = H_{d1}$ (initial length)	2.60	2.60	-	-
$L_1 + E_1 + B_1 = H_{b1}$ (initial)	-	-	3.50	3.50
Lanyard Length at Max Arrest L_2	1.702	2.070	1.702	2.070
Energy Absorber Length At Max Arrest E_2	0.760	0.833	0.842	0.916
Height – Boots to Harness Attachment at Max Arrest B_2	-	-	1.910	1.910
$L_2 + E_2 + D = H_{d2}$ (worst case)	3.062	3.503	-	-
$L_2 + E_2 + B_2 = H_{b2}$ (worst case)	-	-	4.454	4.896
Extension $H_{d2} – H_{d1}$	0.462	0.903	-	-
Extension $H_{b2} – H_{b1}$	-	-	0.954	1.396
At Equilibrium H_e	3.020	3.283	4.412	4.676
Duration at 6kN	118ms	128ms	129ms	139ms



100kg Deadweight, 100kg Human,
3m Free Fall (FF 1.5), 6kN Energy Absorber

Table 6 (ref. Annex B)

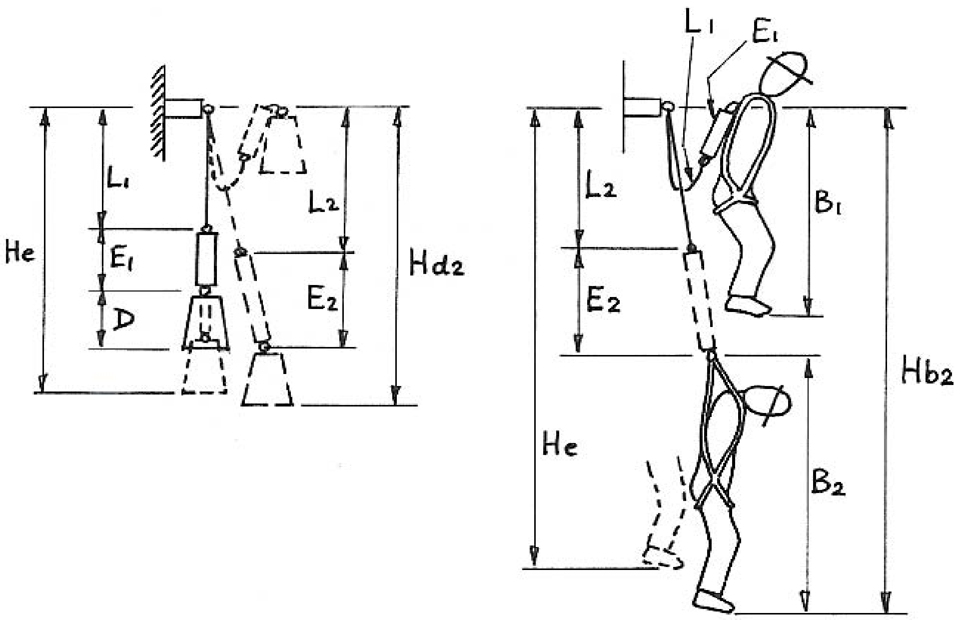
TABLE 6	Webbing Lanyard and Deadweight	Ø12 Nylon Rope Lanyard and Deadweight	Webbing Lanyard and Human	Ø12 Nylon Rope Lanyard and Human
Lanyard Length Including Karabiners L_1	1.65	1.65	1.65	1.65
Energy Absorber Initial Length E_1	0.35	0.35	0.35	0.35
Height of Deadweight D	0.60	0.60	-	-
Height – Boots to Harness Attachment B_1	-	-	1.50	1.50
$L_1 + E_1 + D = H_{d1}$ (initial length)	2.60	2.60	-	-
$L_1 + E_1 + B_1 = H_{b1}$ (initial)	-	-	3.50	3.50
Lanyard Length at Max Arrest L_2	1.702	2.070	1.702	2.070
Energy Absorber Length At Max Arrest E_2	0.960	1.033	1.042	1.115
Height – Boots to Harness Attachment at Max Arrest B_2	-	-	1.910	1.910
$L_2 + E_2 + D = H_{d2}$ (worst case)	3.262	3.703	-	-
$L_2 + E_2 + B_2 = H_{b2}$ (worst case)	-	-	4.654	5.095
Extension $H_{d2} - H_{d1}$	0.662	1.103	-	-
Extension $H_{b2} - H_{b1}$	-	-	1.154	1.595
At Equilibrium H_e	3.220	3.483	4.612	4.875
Duration at 6kN	144ms	152ms	153ms	161ms



100kg Deadweight, 100kg Human,
4m Free Fall (FF 2.0), 6kN Energy Absorber

Table 7 (ref. Annex B)

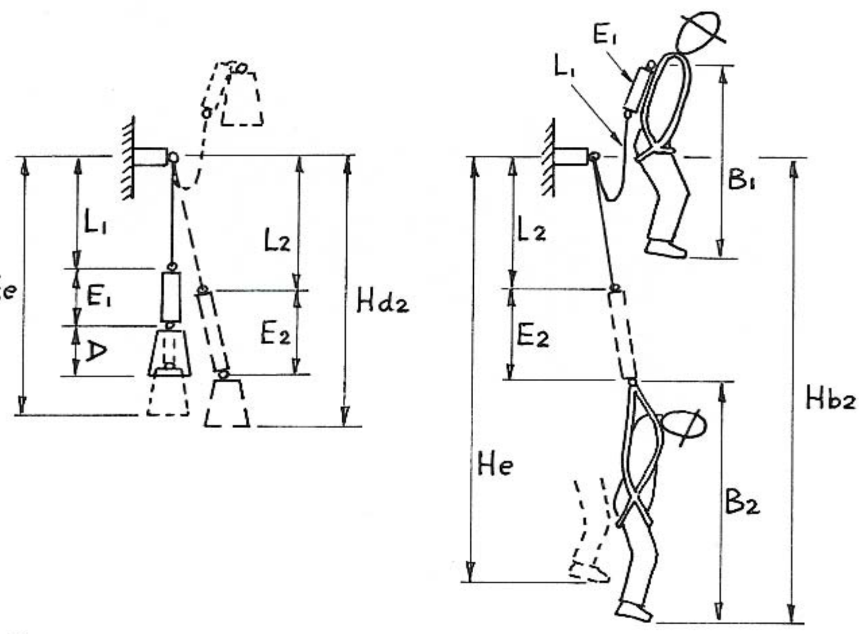
TABLE 7	Webbing Lanyard and Deadweight	Ø12 Nylon Rope Lanyard and Deadweight	Webbing Lanyard and Human	Ø12 Nylon Rope Lanyard and Human
Lanyard Length Including Karabiners L_1	1.65	1.65	1.65	1.65
Energy Absorber Initial Length E_1	0.35	0.35	0.35	0.35
Height of Deadweight D	0.60	0.60	-	-
Height – Boots to Harness Attachment B_1	-	-	1.50	1.50
$L_1 + E_1 + D = H_{d1}$ (initial length)	2.60	2.60	-	-
$L_1 + E_1 + B_1 = H_{b1}$ (initial)	-	-	3.50	3.50
Lanyard Length at Max Arrest L_2	1.702	2.070	1.702	2.070
Energy Absorber Length At Max Arrest E_2	1.160	1.233	1.242	1.315
Height – Boots to Harness Attachment at Max Arrest B_2	-	-	1.910	1.910
$L_2 + E_2 + D = H_{d2}$ (worst case)	3.462	3.903	-	-
$L_2 + E_2 + B_2 = H_{b2}$ (worst case)	-	-	4.854	5.295
Extension $H_{d2} – H_{d1}$	0.862	1.303	-	-
Extension $H_{b2} – H_{b1}$	-	-	1.354	1.795
At Equilibrium H_e	3.420	3.683	4.812	5.075
Duration at 6kN	166ms	173ms	174ms	181ms



100kg Deadweight, 100kg Human,
2m Free Fall (FF 1.0), 8kN Energy Absorber

Table 8 (ref. Annex B)

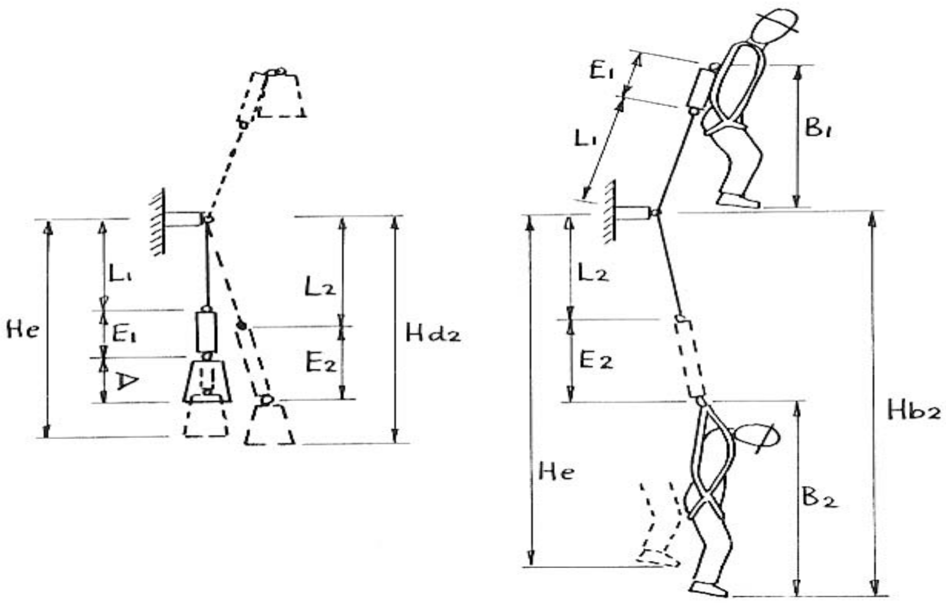
TABLE 8	Webbing Lanyard and Deadweight	Ø12 Nylon Rope Lanyard and Deadweight	Webbing Lanyard and Human	Ø12 Nylon Rope Lanyard and Human
Lanyard Length Including Karabiners L_1	1.65	1.65	1.65	1.65
Energy Absorber Initial Length E_1	0.35	0.35	0.35	0.35
Height of Deadweight D	0.60	0.60	-	-
Height – Boots to Harness Attachment B_1	-	-	1.50	1.50
$L_1 + E_1 + D = H_{d1}$ (initial length)	2.60	2.60	-	-
$L_1 + E_1 + B_1 = H_{b1}$ (initial)	-	-	3.50	3.50
Lanyard Length at Max Arrest L_2	1.710	2.140	1.710	2.140
Energy Absorber Length At Max Arrest E_2	0.644	0.705	0.705	0.766
Height – Boots to Harness Attachment at Max Arrest B_2	-	-	1.930	1.930
$L_2 + E_2 + D = H_{d2}$ (worst case)	2.954	3.445	-	-
$L_2 + E_2 + B_2 = H_{b2}$ (worst case)	-	-	4.345	4.836
Extension $H_{d2} – H_{d1}$	0.354	0.845	-	-
Extension $H_{b2} – H_{b1}$	-	-	0.845	1.336
At Equilibrium H_e	2.904	3.215	4.295	4.606
Duration at 8kN	86ms	95ms	95ms	103ms



100kg Deadweight, 100kg Human,
3m Free Fall (FF 1.5), 8kN Energy Absorber

Table 9 (ref. Annex B)

TABLE 9	Webbing Lanyard and Deadweight	Ø12 Nylon Rope Lanyard and Deadweight	Webbing Lanyard and Human	Ø12 Nylon Rope Lanyard and Human
Lanyard Length Including Karabiners L_1	1.65	1.65	1.65	1.65
Energy Absorber Initial Length E_1	0.35	0.35	0.35	0.35
Height of Deadweight D	0.60	0.60	-	-
Height – Boots to Harness Attachment B_1	-	-	1.50	1.50
$L_1 + E_1 + D = H_{d1}$ (initial length)	2.60	2.60	-	-
$L_1 + E_1 + B_1 = H_{b1}$ (initial)	-	-	3.50	3.50
Lanyard Length at Max Arrest L_2	1.710	2.140	1.710	2.140
Energy Absorber Length At Max Arrest E_2	0.786	0.847	0.848	0.909
Height – Boots to Harness Attachment at Max Arrest B_2	-	-	1.930	1.930
$L_2 + E_2 + D = H_{d2}$ (worst case)	3.096	3.587	-	-
$L_2 + E_2 + B_2 = H_{b2}$ (worst case)	-	-	4.488	4.979
Extension $H_{d2} - H_{d1}$	0.496	0.987	-	-
Extension $H_{b2} - H_{b1}$	-	-	0.988	1.479
At Equilibrium H_e	3.046	3.357	4.438	4.749
Duration at 8kN	105ms	113ms	113ms	119ms



100kg Deadweight, 100kg Human,
4m Free Fall (FF 2.0), 8kN Energy Absorber

Table 10 (ref. Annex B)

TABLE 10	Webbing Lanyard and Deadweight	Ø12 Nylon Rope Lanyard and Deadweight	Webbing Lanyard and Human	Ø12 Nylon Rope Lanyard and Human
Lanyard Length Including Karabiners L_1	1.65	1.65	1.65	1.65
Energy Absorber Initial Length E_1	0.35	0.35	0.35	0.35
Height of Deadweight D	0.60	0.60	-	-
Height – Boots to Harness Attachment B_1	-	-	1.50	1.50
$L_1 + E_1 + D = H_{d1}$ (initial length)	2.60	2.60	-	-
$L_1 + E_1 + B_1 = H_{b1}$ (initial)	-	-	3.50	3.50
Lanyard Length at Max Arrest L_2	1.710	2.140	1.710	2.140
Energy Absorber Length At Max Arrest E_2	0.929	0.990	0.990	1.051
Height – Boots to Harness Attachment at Max Arrest B_2	-	-	1.930	1.930
$L_2 + E_2 + D = H_{d2}$ (worst case)	3.239	3.730	-	-
$L_2 + E_2 + B_2 = H_{b2}$ (worst case)	-	-	4.630	5.121
Extension $H_{d2} – H_{d1}$	0.639	1.130	-	-
Extension $H_{b2} – H_{b1}$	-	-	1.130	1.621
At Equilibrium H_e	3.189	3.500	4.580	4.891
Duration at 8kN	121ms	128ms	128ms	137ms

9 ANNEX C

HIGHLIGHTS OF PAPERS

1 AGARD AR-330, July 1996

Anthropomorphic Dummies for Crash and Escape System Testing, Advisory Group for Aerospace Research & Development (AGARD)

This report provides the history 1949 to 1989 of adult automotive and aerospace anthropomorphic test devices (ATDs) used in the NATO countries since 1949. Included are the Sierra, Grumman Aircraft, Ogle, Hybrid II, Hybrid III, Side Impact Dummy (SID), Advanced Dynamic Anthropomorphic Manikin (ADAM), European Side Impact Dummy (EUROSID 1) and Biofidelic Side Impact Dummy (BIOSID) models.

2 Amphoux, 1991

Physiopathological Aspects of Personal Equipment for Protection Against Falls – Fundamentals of Fall Protection – Edited by Andrew C. Sulowski

Discusses merits of harness attachment points. Disadvantages of pre-sternum (frontal) attachment explained – face of wearer may be struck by lanyard and/or D-ring and possibly more serious danger of whiplash leading to damage of cervical vertebrae. Parachute harness straps (at shoulders) preferable to avoid head/neck movement but these are not practical for industrial users. Dorsal D-ring location is therefore usually most suitable for industrial full body harness. Doctor Amphoux refers to work by Stapp that 12kN is upper limit "already extremely dangerous for a young, well-trained body". He cites Stapp upper limit of $120 m/s^2$ with jolt at $1.3 km/s^3$ (i.e. approximately 12G at 130G/s) for fit servicemen in parachute harness (spine erect, laboratory controlled drop tests). Hence Amphoux is entirely in agreement with 6kN fall arrest limit for unexpected accidental falls in industrial situations, particularly in view of possible 4m (fall factor 2.0 on 2m lanyard) free fall.

3 Amphoux, 1983

Exposure of Human Body in Falling Accidents – International Fall Protection Seminar, Ontario Hydro Research Division

Early parachutists were exposed at canopy opening to forces near 12kN, but parachutists are "young, athletic" and specially trained. For industrial purposes the maxima should be 6kN, though some consider 8kN as maximum. Cites Stapp 5G or 6G over 100ms. Amphoux argues that 6kN is to be preferred because falls are not "symmetric and vertical". Argues that "reduction of the jolt" is desirable but this is future research.

4 ANSI Z359, 1992

Safety Requirements for Personal Fall Arrest Systems, Subsystems and Components – American National Standard

This is a standard written around full body harnesses for users within the weight range 130 to 310 pounds (59 to 140kg). Clause 3, "Requirements", limits the maximum arrest force (MAF) to "not more than 1800 pounds (8.0kN)" and the "deceleration distance" to not more than 42 inches (1,067mm). The drop test (4.2.1.2) specifies a 6ft (1.829m) free fall. It should be noted that the drop test with a self-retracting lanyard is 4ft (1.219m). Informative note E3.1.2 explains that medical research in France in the mid-1970's confirmed earlier USA findings that 12kN is the "threshold" of injury for physically fit personnel subjected to impact "when wearing harnesses" [author's note – i.e. parachute harnesses]. France "arbitrarily" halved this to 6kN in their national standard. Ontario Ministry of Labour adopted 8kN in 1979 and at 1992 there had been no report of death or serious injury associated with the Ontario Provincial standard. Thus 1800lbf (8kN) was considered appropriate for Z359. The specification contains a set of diagrams of harnesses and fall protection systems. Figure 4 in the standard shows various phases of a fall, including the 'rebound' phase, but does not appear to take account of stretch of the harness on a human subject.

5 Ashton-Miller & Schultz, 1988

Biomechanics of the Human Spine and Trunk

This paper has a considerable amount of descriptive material on "skeletal components of the spine and trunk". Actual performances are generally not stated but the reader is referred to the bibliography. A useful description is given of the degeneration of the intervertebral discs "with age the nucleus gradually loses its ability to bind water under mechanical pressure and becomes fibrocartilaginous". On the subject of anthropometry, the paper describes spinal curvature in some detail, lordosis (convex forward) and kyphosis (concave forward) with values normal and abnormal. Tables include range of motion at each spine level and probability of vertebral fatigue failure in the lumbar spine at various load levels. The operation of lumbar trunk muscles is discussed at considerable length along with equations of equilibrium governing these muscles. The authors mention briefly sports injuries including gymnastics, American football, diving, high jumping and butterfly swimmers, but no information is supplied on impact levels.

6 Beeton, Reader, Ernsting 1968

A Personal Torso Parachute Harness and a Modified Restraint Harness for the Type 9 Ejection Seat

Discusses experiments involving high vertical and horizontal accelerations with parachute harness for Martin Baker Type 9 Ejection Seat. Report page 6 states live subjects were dropped from "a predetermined height" – but this height is never stated. Decelerations were within range 5G to 12G. All three subjects claimed to be "comfortable" after tests.

7 Burton et al

Cervical Spinal Injury from Repeated Exposures to Sustained Acceleration, Research and Technology Organisation of NATO (RTO-TR-4)

Chapter 2, Acute Neck Injuries - Burton R R
This report deals mainly with injuries to the neck or cervical spine (the uppermost 7 vertebrae) caused by manoeuvres in the range 4G to 9G. Several pilots suffered bulging cervical discs.

Chapter 8, Pathological and Functional Changes of the Spinal Column in Russian Pilots of High Performance Aircraft - Stupakov et al

The authors make mention that the higher G levels may be an issue for flyers with relatively low density of vertebrae. They also report neck pains occurring in the range 4G to 7.5G whilst the pilots made active head movements lasting several seconds.

Chapter 9, Review of the Vertebral Column Pain Problems in Polish Pilots - Talar et al

This chapter discusses pain to pilots caused by rapid onset-rate (jolt) and high acceleration values (note that Chapter 5.1 speaks of onset rates of 18G/s, whilst 12.4.2 mentions onset rates of 10G/s).

Chapter 12, Biomechanical Considerations in the Development of Cervical Spine Pathologies – Harms-Ringdahl et al

This chapter discusses biomechanical considerations and throws light on the mechanical behaviour of intervertebral discs. From 12.3.1 it is clear that flyers with muscular necks are much less likely to suffer cervical vertebrae damage. On rapid application of force (jolt/impact) the disc behaves as a solid, but with slow applications of force it behaves as a liquid. The compression breaking load (CBL) of cervical vertebrae is similar to that for cervical discs hence, under extreme conditions, either or both will suffer damage with external loading. On the other hand, lumbar discs are approximately 3 times stronger than lumbar vertebrae, thus lumbar vertebrae are more likely to fracture than the disc. The position of the head is important in avoiding cervical spine injury, optimum being the neutral position (erect and facing straight ahead).

Chapter 14. This discusses locations of cervical vertebrae injuries and efforts to improve “neck strength”.

The (RTO-TR-4) bibliographic references run to a total of 138.

8 Chubb R M et al

Compression Fractures of the Spine During USAF Ejections, Aerospace Medicine, October 1965

This is a report of a study of 928 ejections in the years 1960 through 1964. Forty-four individuals suffered compression fractures, 28 during ejection and 16 on parachute landing. Of significance is the conclusion that the erect sitting position, hips and head against the seat, was the most important factor in avoiding ejection injuries.

9 Code C F, et al

Are the Intervertebral Disks Displaced During Positive Acceleration? Aviation Medicine, June 1947

This is a report on measurements and observations obtained by carrying out x-ray examinations of subjects secured in a centrifuge. The centrifugal force direction was head-to-seat, similar to a pilot in an aircraft seat. It was learned that compression up to 6G resulted in reductions of the intervertebral space of no more than 1mm.

10 Crawford H 1980

Proving Tests on Industrial Safety Belts, Harnesses and SafetyLanyards to BS 1397:1979, NEL Report BRAN 02

This report describes a commissioned investigation of the performance of industrial safety belts, harnesses, lanyards and energy absorbers, and compares their performance with the personnel deceleration criteria of BS 1397:1979 'Specification for industrial safety belts, harnesses and safety lanyards'.

11 Deakin 1993

Military Aircrew Head Support System

The author is a researcher with British Aerospace Limited. He describes an ejection seat system that ensures automatic head restraint at the start of an ejection sequence. The background to the project arose from neck injuries to aircrew during violent manoeuvres such as 8G turns. The added weight of sophisticated flying helmets can, when looking around in combat, impose severe "musculoskeletal strain on the neck" leading to incapacitating neck injuries (permanent physiological damage). The problem is compounded if the airman has to eject at 16G, hence the introduction of the military aircrew head support system (MAHSS).

12 Delahaye 1970

Physiopathology and Pathology of Affections of the Spine in Aerospace Medicine, Advisory Group for Aerospace Research & Development (AGARD) – AGARD-AG-140-70

Sources of this work are predominantly French. Acceleration/time diagrams are modifications of the Stapp/Webb/Eiband data presented in NASA Bioastronautics Data Book. Intervertebral discs and their nucleus pulposis (central pulp material) are described at 3.2. Ejection accelerations "greater than 15g" during 200–400ms, and parachute canopy opening shocks of 8-10g for 1.0s are mentioned at 2.2.1 (and page 24 describes several ejection systems of 20-21g for 80-100ms, also 16-18g for 180-200ms). At page 10 the strength of vertebrae is discussed, with "creakings, compression and extravasation" of blood at 600–700kg, and rupture at "about 850kg". Muscle support must also be taken into account, so the vertebrae can probably tolerate compression of "one ton". Page 20 of the report illustrates force/time traces for ejection seat systems, one of them being for 16g over some 450ms. Text and graphs, pages 30 to 33 highlight the fact that decelerations measured at the seat are graduated as the compression forces are distributed up and along the vertebrae. The bibliography runs to 219 entries.

13 Delahaye et al, 1982

Physiopathology and Pathology of Affections of the Spine in Aerospace Medicine (Second Edition), Advisory Group for Aerospace Research & Development (AGARD) – AGARD-AG-250(Eng), edited by Delahaye R P and Auffret R

This is a document of 335 pages dealing mainly with the significance of spinal disorders in aerospace medicine. In discrete chapters it deals with anatomy of the spine, biomechanics of the spine, spinal stresses in flight, traumatic lesions of the spine, postural disorders, the spine and fitness for flight and medico-legal aspects. Much of the subjects of anatomy and biomechanics have been repeated from the previous edition (AGARD-AG-140-70), with observations:

Chapter 2, Anatomy of the Spine – Kleitz and Delahaye
The height of discs slowly decreases from the cervical column (discs 2mm to 4mm) downwards to the 5th thoracic vertebra, and increases again down the spine to reach the greatest dimension between L4 and L5 (average 12mm), then decreases again. Some 50 to 60% of the cross section of a disc comprises the nucleus pulposus which is 90 to 75% water depending on age. Average length of the spinal column is 75cm, 30% to 35% composed of discs.

Chapter 3, Biomechanics of the Spine – Kleitz and Delahaye
Compression tests on the lumbar column have proved that the vertebrae fracture before any rupture of the discs. The trunk and back muscles add to stability. However, clinical observations indicate degeneration and damage to discs are more frequent beyond 50 years of age.

Chapter 4, Spinal Stresses in Flight – Auffret and Viellefond
This chapter includes modifications of the Eiband/Stapp acceleration figures as they relate to ejection seats. It includes strong comments about the need to take account of the rate of onset of acceleration (jolt). It defines impact as high acceleration with duration of application less than 0.2 seconds. Ejection necessarily requires high $+G_z$ forces (>15G) with duration 0.2 to 0.5 seconds. Opening shock of an ejection system parachute is of the order of 8G to 10G for one second, whereas ground impact tends to be 2G to 5G with duration 0.1 to 0.4 seconds.

Chapter 5, Traumatic Lesions of the Spine in Aviation Medicine –Delahaye and Metges
Attention is drawn to position of the head at $+G_z$ arrest so as to avoid damage to the cervical spine. This chapter describes hyperextension – the condition where, at deceleration, head movements front-to-back and back-to-front (the classic "bell-ringing" motion) can be more serious and more rapid than "whiplash". The statistical work of Teyssandier for parachute jumps by French airborne troops is shown in graphical form at figure 71. Theoretical aspects of parachute opening are covered in this chapter. Depending upon the canopy opening, forces lie between 5kN and 8kN; a peak of 12kN is described as the "currently accepted maximum of tolerance". Further included in this chapter, for interest, are landing conditions for 'parawing' parachute users ("landing shock" varies "with the square of the horizontal wind speed") and the observation that high wind speeds account for a high proportion of landing injuries. There is a short section (p 125) on "impacts at terminal velocity" into rice fields and snow (including the classic RAF Lancaster tail gunner who survived without a parachute - the writers state, "this Scotsman jumped without it from a height of 5500m").

14 Eiband A M 1959

Human Tolerance to Rapidly Applied Accelerations, A Summary of the Literature, NASA Memorandum 5-19-59E

This is a 'classic' paper with data drawn from many reference sources. The writer begins by pointing out that the results of his work "indicate that adequate torso and extremity restraint is the primary variable in tolerance to rapidly applied accelerations" and that the harness must "transmit the major portion of the accelerating force directly to the pelvic structure and not via the vertebral column". Eiband further stresses the importance of onset rate, magnitude and duration of accelerating force. The discussion of terminology for direction of forces is thorough. The paper seeks to identify from animal (hogs and chimpanzee) and human experiments the separating impact bands for high risk, medium risk and low risk deceleration (for spineward/sternward and headward/tailward) forces. (Eiband's plotted representations of these data have featured in many later studies of biomechanics by other investigators.)

15 Ernsting, 1967, Farnborough

10 'g' Deceleration Drop Tests of Subjects Wearing the Phantom AEAS

Martin Baker Type 9 harness with SARBE Mk2 life saving waistcoat. 1.2m (4ft) drops, three subjects. 12 total drops with three subjects, 8 drops no record of deceleration, remaining 4 were 9G to 10G with two subjects. Both "comfortable" following tests.

16 Ewing & Thomas

Human Dynamic Response to $-G_x$ Impact Acceleration – AGARD-CP-88-71, Linear Acceleration of Impact Type, Ref. 11

This is a report on arrest of forward motion, i.e. $-G_x$ acceleration. Although fore/aft accelerations are not the primary subject of this study (authors note, HSE study), the indications of the AGARD paper should be noted as "all accelerations acting on the head and neck must be transmitted through the vertebral column". The paper illustrates in graph form accelerations at the level of the head compared with spine resultant accelerations. The data are gathered from sled runs of nominal 10G deceleration at 250G/s rate of onset.

17 Fraser, NASA 1973

Sustained Linear Acceleration

Discussion of physiological effects of sustained acceleration, effects on blood pressure and time to unconsciousness as a function of rate of onset of positive acceleration ($+G_z$). Discussion of resonant frequency of human body sitting erect. Discusses also vision, body motion, control and mental function under sustained acceleration.

18 Glaister, 1978, Farnborough

Human Tolerance to Impact Acceleration

This paper makes the observation that a pilot seated on an ejection seat has a natural frequency of approximately 5Hz, "so the critical pulse length is about 0.2s". However, "if the peak forces are kept below injurious levels the duration of exposure becomes less important". The paper is an important discussion of the effect of velocity change up to 0.25s period, and of pulse duration as a fraction of the natural period of the subject.

19 Guignard, 1961, Farnborough

Result of a Resonance Search Test on the RAE Prototype Anthropomorphic Parachute Test Dummy

Test showed dummy vertebral column to have 3 cycles/s (3Hz) resonant frequency. Compares with 5Hz obtained from live men in paper by Guignard and Irving, 1960.

20 Hearon, et al

Mechanism of Vertebral Fracture in the F/FB-111 Ejection Experience

This is a review of accident investigation reports of all non-fatal F/FB-111 ejections October 1967 to March 1980. Spinal injuries occurred in 23 of the 78 cases investigated. The general mechanism of injury involved both axial compression and flexion. Ejection seat force levels are not given in this paper, but it is probable that these were similar to the levels indicated in the Eiband/Stapp data.

21 Hearon & Brinkley 1984

Fall Arrest and Post-fall suspension: Literature Review and Directions for Further Research

Although named second in the authorship of this paper, James Brinkley was one of the leading communicators of 'available' US Airforce research in the 1980s, when interest in controlled fall-arrest was quickening. The paper reviews drop tests carried out on anaesthetised dogs, it also makes mention of tests carried out on human subjects in France by Amphoux, Ardouin and Noel. It also describes drop tests using a waist belt using shoulder straps. Maurice Amphoux suffered two fractured ribs on a fall of 0.5m and a stuntman declined further tests following thoracic contusions on a fall of 0.8m. A drop test of 2m (FF 1.0) with a human subject in a full-body harness with energy absorber was recorded by Ulysse. The arrest force was 3.3G. Work by Beeton et al (q.v.) in the UK, with human subjects in parachute harness (5G to 12G) is also described. Brinkley ends the paper with the suggestion for further research "future efforts should be based on the scientific method, beginning with the formulation of hypotheses amenable to experimental evaluation".

22 Henzel J H, 1967

The Human Spinal Column And Upward Ejection Acceleration: An Appraisal of Biodynamic Implications

Henzel begins with a review of "aircraft – pilot separation". His short history begins with German work prior to World War II and the first installation of an ejection seat by Junkers in 1941. Heinkel and Dornier were also involved in research. Early German experience was concerned with cervical injury, "snapping" of the head, neck and shoulders. He then reviews the work of the Martin Baker company circa 1944 and Swedish work 1946. Development of ejection systems in the USA began in 1945 and was accompanied by a comprehensive "aeromedical" phase. Both the engineering and aeromedical phases were aimed at development of a system to safely clear a man from an aircraft travelling at 600mph. This work was later developed to permit clearance from an aircraft "in trouble" before it left the ground.

The report deals primarily with the vertebral body and intervertebral disc behaviour during ejection. Henzel points out that many "ejection-incurred spinal injuries result from abnormal ejectee posture". Discussing physiology of the vertebral column, he states "When an adult human male moves from a reclining to an upright position, lower vertebral column disc nuclei are subjected to loads that average 45kg. If this same individual bends or extends his spine, as is often done when one stretches, this same nucleus must support from 100 to 130kg. When the body is bent forward at a 90º angle, the pressure transmitted to the lower lumbar discs is about ten times as great as the weight being lifted or supported".

"The implications of annular strain and vertebral body loading occurring under these conditions and in the presence of a degenerated disc are apparent. A poorly postured spine undergoing accelerative forces experiences the same type of load distribution, but in a dynamic manner".

Henzel reviews the work of Ruff, Stech and Perey and their tests on vertebrae strength, particularly with respect to end-plate damage (end-plate as in top and base of a metal can). In tabular and graph form he illustrates the stages of compression testing as end-plate fracture, limit of proportionality, yield point and compression fracture. He relates the modern test findings to those of Jefferson (1928) and the observation that then, as now, the most vulnerable regions of the spine were C5 to C7 and T12 and L1. Jefferson's work covered 2006 cases of spinal fracture.

23 Higgins L S et al, 1965

Studies on Vertebral Injuries Sustained During Aircrew Ejection

This report was prepared for the US Office of Naval Research. It begins with comments on a comprehensive survey of available world literature on vertebral injuries to aviators on ejection. These cover research in Germany during World War II, Martin-Baker studies 1944 to 1946, Swedish developments 1945 and US work 1945 to 1946. The early German work was conducted by Siegfried Ruff whose experiments included the breaking load of vertebrae under pressure and compressibility of the intervertebral disc. Ruff determined that a column of 6 vertebrae (T-10 through L-3) withstood 690kg (6.75kN), at which T12 ruptured. On a separate test of 7 lower vertebrae, T-8 failed at 540kg (5.28kN).

The paper summarises Ruff's early work on individual vertebrae (based on body weight of 75kg) as follows:

Vertebrae	*Compressive strength (kgf)*	*% of body weigth borne*
T-8	540-640	33
T-9	610-720	37
T-10	660-800	40
T-11	720-860	44
T-12	690-900	47
L-1	720-900	50
L-2	800-990	53
L-3	900-1100	56
L-4	900-1200	58
L-5	1000	60

No information is given as to the terminal age of the cadavers, but Olof Perey of Sweden conducted tests on the strength of the lumbar vertebrae and found that the average breaking strength for specimens over 60 years of age was 425kgf whilst, for those under 40 years of age, the average strength was 800kgf.

Brown et al of Massachusetts General Hospital and MIT found the following (ages are not given):

Body	*Vertebrae*	*Compressive strength (kgf)*
A	L-2, L-3	500
	L-4, L-5	455
B	L-3, L-4	545
	L-5, S-1	570
C	L-4, L-5	590

Evans et al at Wayne State University had results as follows:

Vertebrae	*Compressive strength (kgf)*
T12 – L1	364 (max 709)
L1 – L2	361 (max 679)
L2 – L3	420 (max 898)
L3 –L4	425 (max 684)
L4 –L5	402 (max 754)
L5 - sacrum	351 (max 563)

Only Ruff is quoted as providing information on relevant body weight, and only Perey is quoted on the influence of age.

24 Iatridis J C et al, 1996

Is the Nucleus Pulposus a Solid or a Fluid? Mechanical Behaviours of the Nucleus Pulposus of the Human Intervertebral Disc

This study concentrated on the behaviour of the nucleus pulposus in torsional shear under transient and dynamic conditions. The conclusion of the study was that at slow deformation rates the shear stress of the nucleus pulposus relaxed nearly to zero, indicating "fluid-like" nature of the tissue. Under dynamic conditions it had a predominantly "solid-like" nature similar to biological solids.

25 Jones W L et al

Ejection Seat Accelerations and Injuries, Aerospace Medicine, June 1964

This is a report on the Martin-Baker Ejection Escape System as fitted in the USA Navy aircraft, Grumann Jet Trainer F-9J. Studies by Grumann showed that, with 95% confidence, the seat acceleration at 100kts was within 17.7G to 21.7G, and at 400kts it was 18.3G to 22.2G. The onset rates varied from 180G/s to 383G/s. It was suspected that the wide variation of ejection accelerations explained a number of vertebral fractures to some users of the system and no injury to other pilots using the same ejection equipment. The Naval Ordnance department developed a rocket assisted catapult system with a first phase of duration 200ms peaking at 11.8G and a second phase of 400ms at 6.5G. It is suspected that the most serious injuries with the Martin-Baker equipment "were probably in the neighbourhood of 20G or greater".

26 Kazarian et al

Biomechanics of the Vertebral Column and Internal Organ Response to Seated Spinal Impact in the Rhesus Monkey (Macaca Mulatta), Aerospace Medical Research Laboratory, document AMRL-TR-70-85

Rhesus monkeys were anaesthetised and restrained by lap belt and torso harness in a vertical impact carriage. They were exposed to $+G_z$ rectangular deceleration/time impacts in the range 25 to 900G and duration 2 to 22ms. Scaling factors were used to relate the results to human situations. The results agree with other researchers that there are two factors determining injury, i.e. rate of onset and peak G. Injuries to the monkeys due to severity of impact ranged from lung damage at around 90G (measured at seat level) and above, spinal damage at around 100G and above, liver damage at around 140G and above, and heart damage at about 160G and above. The rectangular impact events causing these injuries were of the above quoted durations. These results were compared with the US military specification 9479A 5% probability of injury level of 12.1G. The authors were of the opinion that, adequately constrained, the "human body could withstand input accelerations of up to 20G".

27 Kazarian et al, 1971

The Dynamic Biomechanical Nature of Spinal Fractures and Articular Facet Derangement

This is a study of the impact performance of anaesthetised rhesus monkeys exposed to $+G_z$ seated acceleration of square form (i.e. rapid rate of onset). The results for vertebral damage are presented in bar chart form and compared with vertebral injuries in aircrew surviving ejection. In graph form the 99% probability of injury (lung, vertebrae, liver and heart) for the rhesus monkey is compared with the 12.1G, 5% probability of injury for man based on US Mil-Spec-9497A-Code 2.

28 Norris, 1974, et al, 1971

Preliminary Investigation into the Hanging Characteristics of the Lightning MK 3 AEA when Supported in a Combined Harness Assembly of a Martin Baker MK 4 BSC Seat

Suspension tests only – crotch pain, severe testicular.

29 Orzech et al 1987

Evaluation of Full-body Harnesses During Prolonged Motionless Suspension of Volunteers

As the title shows, this paper also deals with suspension tests only but is useful, among other things for its photographs and details of a number of full-body harness designs. It also describes life-threatening symptoms of suspended human subjects and hence highlights the reason for rapid rescue after a fall.

30 OSHA – 29 CFR

Personal Fall Arrest Systems (PFAS) – 1915.159, Occupational Safety and Health Administration, US Department of Labor

Clause (b)(6)(ii) limits the arrest force to 8kN with a "body harness" and (b)(6)(iii) limits the deceleration distance to 3.5 feet (1.07m). Clause 159(b)7 states that the system must be such that the wearer cannot fall more than 6ft (1.8m) nor hit a lower level.

31 Reader, 1967, Farnborough

Measurement of Loads on Combined Life Saving Waistcoat and Torso Harness Closure Plate During Simulated Man-Seat Separation and Parachute Deployment

Test results with 2 subjects. Drop heights 150mm (6"), 300mm (12"), 600mm (24"), 900mm (36"), 1.2m (48") in range 3G to 9G, no injuries, slight crotch discomfort. On drops 300mm, 600mm and 1.9m [4.2-6.5G] with asymmetric suspension there was considerable discomfort in groin.

32 Reader, 1968, Farnborough

An Assessment of Industrial Safety Harnesses

Eight live subjects in weight range 20 to 98 percentile, with 9 harnesses and 3 waist belts. Only 3 harnesses were drop tested, remaining 6 were uncomfortable or had dangerous features. Waist belts were not drop tested. 10ft steel cable, 2ft drops, 4G to 4.6G.

33 Reader, 1979, Farnborough

A New Safety Harness for Mobile Aircrew

Suspension tests only, with live personnel. Drop tests with OGLE 95th percentile dummy, 1.14m drop produced 5.3G, 1.5m drop produced 5.7G, 2m drop produced 7G, 3m drop 9.7G. All carried out with chest harness life preserver.

34 Reader, 1970, Farnborough

The Load Distribution in an Ejection Seat Combined Harness Under Simulated Parachute Canopy Inflation

Interesting explanation that, due to reduced air density, terminal velocity is greater at high altitude. Thus a parachute canopy inflates more rapidly at high altitude and deceleration forces on the airman are greater than with ejection at low altitude. A SIERRA 50th percentile dummy was used for dynamic tests on 4 ejection seat parachute harnesses. All four harnesses survived 20G (with onset 214 g/sec, and 125ms duration).

35 Reader, 1969, Farnborough

An Assessment of a Lightweight Constant Wear Harness to Replace the D MK.1 Harness

Eight live subjects for suspension tests, in range 109lb [1 percentile] to 221lb [99 percentile]. Four live subjects for dynamic tests, 3 drop heights each.

Subject	*Percentile*	*Force at 300mm drop height*	*Force at 600mm drop height*	*Force at 1.05m drop height*
Subject A	87th percentile	3.3G	6.2G	9.0G
Subject B	10th percentile	N/R	7.2G	9.9G
Subject C	47th percentile	2.5G	5.0G	8.8G
Subject D	55th percentile	4.7G	7.2G	8.5G

This parachute harness was fitted with shoulder straps. None of the subjects complained of back pain despite the 8.5G to 9.9G arrest forces, but subjects B and C did report discomfort at groin and under buttocks.

36 Sances et al 1981

Bioengineering Analysis of Head and Spine Injuries

The early parts of this paper deal largely with head and spinal injuries deriving from blows through the head to the spine. In this sense much of the paper is outwith the realms of the present study. Where the paper does deal with impact tolerance the authors cite the work of Snyder, Stapp, Ruff and Ewing.

The paper is especially interesting in illustrating the variation in strength of vertebrae and intervertebral discs through the spinal column, as follows:

Vertebrae	*Region*	*Compressive breaking force (N)*
	Cervical	3089
	Upper thoracic	3020
	Lower thoracic	4491
	Lumbar	4952

Intervertebral disc	*Region*	*Compressive breaking force (N)*
	Cervical	3138
	Upper thoracic	4413
	Lower thoracic	11278
	Lumbar	14710

The authors also quote work by Nachemson to demonstrate the forces in the 3rd lumbar disc in different positions of individuals of body weight 50, 70 and 100kg:

Position of Body	*Force for50kg person (N)*	*Force for70kg person (N)*	*Force for 100kg person (N)*
Upright sitting, unsupported	1079N	1392	1863
Upright standing	735	971	1324
Reclining, supine	147	196	245
Sitting + flexion 20°	1422	1873	2550
Sitting + flexion 20° and 10kg each hand	2216	2648	3334
Standing + flexion 20°	1079	1451	2010
Standing + flexion 20° and 10kg each hand	1736	2109	2815

Note - to the layman it seems illogical that the lumbar disc should experience a compressive force greater than the weight of the trunk above it. White and Panjabi explain at page 56 of their work that the centre of gravity of the trunk is in front of the spine, causing a bending moment. The muscles of the trunk counteract the bending moment and give rise to large compressive disc forces.

37 Schall D G

Non-Ejection Cervical Spine Injuries Due to +Gz in High Performance Aircraft

This paper opens with the interesting observation that native Africans frequently carry heavy loads on their heads, and that the cervical spine can sustain "axial loads up to 91kg without difficulty". The major concern of the paper is that pilots of high performance aircraft wearing helmets can be submitted to forces of up to 65kgf (636N) in $+9G_z$ manoeuvres and, if the pilot is not in an 'upright' position, non-axial forces can cause injury to the head-neck complex due to flexion. Eight such case histories are described. In addition to techniques for improving head-positioning in the cockpit, the author suggests neck strengthening exercises (such as are recommended for American football players) as an aid to reducing such injuries.

38 Seddon P

Harness suspension: review and evaluation of existing information, HSE Contract Research Report 451/2002

This review considers the suspension phase of a fall. The body may be in a state of shock, badly injured or unconscious, with little likelihood of movement of the legs. The paper reviews research papers on suspension trauma published between 1968 and 1998. Eleven harness standards are reviewed, revealing "surprising lack of attention to the suspension phase of a fall and its inherent dangers".

39 Shaw R S

Ruptured Intervertebral Disc from Positive Acceleration, Aviation Medicine, August 1948

This paper relates to spinal loading experienced by dive-bomber pilots. The paper mentions "early German work" with ejection seats where subjects suffered compression fractures of lumbar and thoracic vertebrae from accelerations of 25G to 35G. Static testing of fresh "human spines" has (at 1948) demonstrated failure at about 20G. The author discusses cases of back pain at 5G and herniated nucleus pulposus (hernia of centre of intervertebral disc) at 7G to 9G. The conclusion drawn from the discussion is that these pilots had "an awkward, flexed position of the back" during flight pull-out, and thus suffered injuries "from accelerations ordinarily considered well below the tolerance limit for back injury". The paper concludes that a flexed spine can be injured much more readily than an erect spine.

40 Snyder, 1973

IMPACT, Chapter 6 – Bioastronautics Data Book, Second Edition, 1973 NASA SP-3006

Agrees with Stapp (1961) that impact involves forces of up to 0.2 sec duration. Continues, "knowledge concerning human impact tolerances is very incomplete", and volunteers mainly "young healthy male subjects under rigidly controlled conditions". Cadavers and animals have also been used. Full explanation of the physiological standard recommended by the Biodynamics Committee of the Aerospace Medical Panel, AGARD, for uniaxial accelerations, also vernacular 'eyeballs' movement as inertial response to applied acceleration, e.g.

upward acceleration = positive G = $+G_z$ = eyeballs down.

Rise time/jerk/jolt important. Pages 231 to 233 are important in discussing, and showing by graph, "survivable, abrupt, vertical ($\pm G_z$) impact". Table 6-10, pages 260 to 264, deals with catapult and tower ejection seat tests. Item 13, therein, states "injury at >12G if head not held erect". This chapter warrants much more analysis.

41 Snyder

Man's Survivability of Extreme Forces in Free-fall Impact – AGARD-CP-88-71, Linear Acceleration of Impact Type, Ref. 5

Review of survivals of falls onto steel, concrete, soil, water and snow. Falls, even falls from aircraft, into snow have provided "greatest impact energy absorption and incidence of survival". The writer highlights (on physical aspects of fall arrest) rate of onset, magnitude of force, impacted material, direction of force and its distribution. Biological aspects include age, sex, physical condition, mental condition, tissue properties and secondary impacts. Rate of onset is seen to be very important, as is total duration of the force. The writer quotes Stapp's suggestion that when impact duration "is less than 0.2 seconds, the tissues react with damage to structural integrity, behaving like inert materials under conditions of mechanical stress analysis, where structural damage and failure are independent of gradients of fluid displacement". (Attention is thus drawn to damage to internal organs in impact trauma). The writer points out that, at that time (1971), there were insufficient data on how force and deformation vary with time in ultra-short impacts. There is a strong correlation between fitness condition and impact tolerance, the best survival rates being among young, athletic males and females. Muscle relaxation and effects of alcohol and drugs can often be a factor. Discussing falls into snow, the writer cites accounts of the German 4th Army who witnessed the Russian Yukhnov airborne operation of 1942 where "in one instance the airborne troops were placed in sacks filled with straw and dropped without parachutes into snow". During the Russian-Finish war the Soviets "experimented with dropping troops without parachutes into snow from heights of 15 to 50 feet". It is reported that "some 50% of the men were dropped successfully".

42 Snyder R G 1963

Human Tolerances to Extreme Impacts in Free-Fall, Aerospace Medicine Volume 34, No.8, August 1963

As the title indicates, this paper deals with falls from 'great height'. Several extraordinary fall-accounts are given and the writer in effect agrees with Stapp when stating "it is possible that human tissues may react differently to impact energy if subjected to a high enough force in an extremely short time period". Most of this paper is outwith the bounds of the present HSE sponsored study but Snyder, when discussing $+G_z$ 'seated' impacts, states that "pelvic and vertebral trauma are prevalent . . particularly in the L-4 to T-12 area". He further comments that "supporting tissue structures . . . are often damaged in impact but are not diagnosed due to more painful complications masking such injuries".

43 Stapp J P 1961

Human Tolerance to Severe, Abrupt Acceleration

Colonel J P Stapp is internationally known in the sphere of biodynamics. He discusses investigations of tolerance beginning with World War II in Germany, describing a "guillotine type" of drop test and 20G peak decelerations causing back pain. Other investigations on a range of cadaver vertebrae identified static test compression fractures of 540kg for the 8th thoracic and 4200kg for 5th lumbar. In dynamic tests the first lumbar vertebrae withstood 18G to 23G. Following World War II several rocket driven sleds were built in the USA for study of "tolerance and survival values" for "magnitude, duration and rate of change of negative acceleration". Stapp points out that centrifuge experiments are limited by the slow rate of onset. Of significance is the argument that duration of accelerative forces less than 0.2 seconds causes body tissues to "behave like inert materials under conditions of mechanical stress analysis". He illustrates the paper with the Eiband (1959) graphs of duration and magnitude of acceleration in the various directions, spineward, sternumward, headward and tailward. He also includes damage sensitivity curves from Kornhauser, 1958. In the discussion of "abrupt headward acceleration (positive G)", the writer states that with a vertebral column of correct alignment and restraint, "35G can be tolerated at an onset rate of 500G per second or less". He quotes other papers which state that a maximum rate of onset is 1300G per second, but the corresponding force is 12G maximum. He stresses that "body support is critical in this orientation". Poor body support with hyperflexion can lead to serious damage of the lumbar spine. Citing Eiband, spinal injury has been caused at forces between 3G and 4G when the subject is restrained only with a lap strap.

44 States J D

Biodynamics of Sports Injuries – AGARD-CP-88-71, Linear Acceleration of Impact Type, Ref. 8

John D States, MD, Assistant Clinical Professor of Surgery (Orthopaedic), University of Rochester School of Medicine, describes sports and recreational injuries and fatalities including US football, auto racing, mountain climbing and sky diving. The paper is largely statistical. It does not include any estimates of forces causing the injuries.

45 Steinberg 1977

A Study of Personal Fall-Safety Equipment

This study was undertaken for the Occupational Safety and Health Administration (OSHA) of the USA. At the time of its writing OSHA considered that standards for fall-safety systems were inadequate. Steinberg reviewed body belts, body harnesses, impact accelerations and forces, lanyards and fall arrest equipment in its widest sense. He draws his information from the Eiband curves and from the aviation medicine researches of Shaw and the Naval Aerospace Medical Research Laboratory.

From these data he recommends that, with a body belt, the ‘shock load’ be “limited to $8g_n$” with a maximum free-fall of 600mm (2ft). For a “body harness” a maximum free-fall of 1.8m (6ft) is recommended. Commenting on the then available American National Standards Institute (ANSI) and the Canadian Standards Association (CSA) adoption of “a $10g_n$ acceleration limit in the $+a_z$ direction”, he states this limit “may be higher than should be allowed when body orientation can not be controlled and especially with the use of body belts”. His reason for this conclusion is based on evaluation of ejection seat acceleration “after separation from the aircraft, the US Navy limits $+a_z$ to $17g_n$ for an expected rate of spinal injury of 5% or less” (author's note, presumably $17g_n$ 'in-seat'). Steinberg quotes from the work of a certain CT Morgan and his researches into ejection seat/parachute injuries at high altitudes. Morgan stated “parachutes opening shocks are greatest at high altitudes, and that impact accelerations below $20g_n$ are considered safe, 20 to $30g_n$ are borderline and over $30g_n$ are dangerous for man, parachute and harness”. He then cites work on “body belts” done by Ardouin and a French medical team whose conclusion was “it would be an exceptional person that could withstand accelerations greater than 6 to $8g_n$”. He also cites work by the German Alpine Club and their conclusion that falls in “waste tie-ins” can result in death when forces exceed 3.75kN (or 3.75G). In addition to the above, Steinberg makes a useful observation concerning use of a “lineman’s belt”, that “there is a high probability that a fall would result in a $+a_x$ acceleration and backbend or reverse jackknife”, and that “the acceptable level of average body acceleration is probably only 4 to $5g_n$”.

Much of the paper is taken up with descriptions of spring systems and related formulae. There are useful tables of percentile body weights for males by age in the USA and for construction workers when clothed and equipped. He observes that accepted forces for aircrew may incur vertebral injury, and cautions “It is most unlikely that a construction worker falling into a body harness-lanyard system (even one with riser straps off each shoulder) will impact in an optimal configuration”. Hence Steinberg advises OSHA that $8g_n$ should be considered as maximum on a 1.8m (6ft) fall using full body harness and a 1.8m long lanyard.

46 Stupakov G P and Mazurin Yu V

The Limits of Human Impact Acceleration Tolerance, AIAA paper no. 93-3572

The abstract for this paper reads: "The paper discusses experimental and theoretical aspects of grounding of new calculated criteria of trauma-safety and subjectively estimated by man the effect of impact acceleration exposure, considering the factors of pulse time duration enhancement and posture change, which are characteristic for modern ejection seats in high maneuverable aircrafts". The received microfiche copy from the British Library was photographically overexposed and the graphs were impossible to read, but it is clear from the text that the subject of the paper relates to "modern ejection seats". The paper describes centrifuge training of test subjects which resulted in average elevation of bone tissue density of 4%.

47 Swearingen J J et al

Human Voluntary Tolerance to Vertical Impact, Aerospace Medicine, December 1960

The writer acknowledges the intensive study of horizontal accelerations by Stapp and others and points out that this study concentrates on human subjects at high vertical impact forces (positive G), high jolt factors and short durations. The positions considered were seated, standing with legs rigid, knees bent, and finally squatting. 13 human subjects were involved in the test programme of nearly 500 drop tests. Describing seated drop tests (without back support) Swearingen reports that 95G with onset rate of 19,000G/s caused shoulder mounted accelerometers to record 10G with rate of onset 600G/s. The subjects survived the tests but all complained of severe pains in the chest, stomach, lower spine and head. He then suggests that these levels be treated as the safe limit for maximum voluntary tolerance "in view of the fact that helicopter pilots have died of ruptured aortas in vertical crashes and that the etiology of not only the chest pains but also the severe pains in the stomach, spine and top of the head are unknown". In tests with standing subjects, knees locked, 65G at 10,000G/s (at feet level) caused 10G at 600G/s at shoulder level. Reported pain was similar to the seated tests. The height of the test equipment limited the tests with knees bent to 250G, 50,000G/s at the platform and 7G, 583G/s at the shoulders.

The pain levels were not quite so high but the deceleration forces overcame leg muscle strength and all of the subjects ended in a squat position. The squatting position tests had an input of 133G at 26,600G/s causing shoulder readings of 5G at 250G/s. The subjects recorded no pains in the head or trunk but all reported severe pains in the knee area. The force/ time traces for the 4 test conditions all indicate some bouncing of the drop platform with peaks as described above. The platform impact event (with bounces) was completed within 14ms for the seated tests, 24ms for the standing – knees locked tests, 15ms for knees bent and 19ms for squatting position. The readings at shoulder level indicate durations of 65ms for the seated position, 80ms for the standing knees locked position, 160ms for knees bent and 200ms for the squat position.

48 Teyssandier 1967

Les Atteintes Traumatiques Du Rachis Chez Le Parachutiste [Traumatic Injuries of the Spine for the Paratrooper]

Statistical data on 1,033,525 jumps during 1959 to 1966 at the school for paratroopers at Pau, France, recording 947 traumatic injuries. Line of force on landing is vector through 12th dorsal to 3rd lumbar. Writer comments on work by Decoulx and Rieunau who demonstrated that, to crush 3 sound lumbar vertebrae requires vertical force in order of 600 to 800kgf [i.e. 6kN to 8kN]. Writer concludes traumatic injury of the spine of 0.1% [1 per 1,000] of jumps for paratroopers represents 10% of total accidents and 50% of total breaks during the 1,033,525 jumps. Report includes 'classic' diagrams of spinal column configuration at parachute opening and on land

49 Teyssandier 1967

Les Fractures Du Rachis Chez Le Parachutiste [Fractures of the Spine for the Paratrooper]

Survey of 1,468,399 jumps, recording 219 fractures of the spine i.e. 0.0149% [1.49 per 10,000] jumps. The writer comments that fractures of the cervical vertebrae [region of neck] were numerous with the French type T5 parachute folded to open "sail first". Developments in parachute design and folding procedures have reduced the causal opening shock and breaks of the cervical vertebrae were now rare. Of the recorded spinal injuries, 76% occurred in the lower back between 12th dorsal and 3rd lumbar vertebrae. First lumbar vertebra damage is the most frequent injury, 35%. Extensive French language bibliography may warrant further study.

50 Virgin W J

Experimental Investigations into the Physical Properties of the Intervertebral Disc, The Journal of Bone and Joint Surgery, Vol.33B, No.4, November 1951

This is a report on the physical strength of intervertebral discs removed from 51 cadavers. The writer describes the intervertebral disc "mechanically" as a "viscous elastic structure" generally with a "modulus increasing with load", unlike metals where load and extension are proportional. Load/deflection tests showed hysteresis to be smaller in the upper lumbar and lower dorsal discs, largest in the lower lumbar discs. In very young subjects the hysteresis was "very large". "It was least in people in the middle decade of life" and in elderly where there was no degeneration in the discs. The writer argues that efficiency of the disc improves with repeated loading, significant for people doing heavy work, e.g. the custom of "taking the strain" before lifting a heavy load. He also concludes that the intervertebral disc reaches greatest efficiency in adult life.

51 Wallace & Swearingen 1971

Crash Injury Severity as Related to Aircraft Attitude During Impact

This paper is a review of "nearly 100 selected general aviation accidents". The purpose of the paper was to demonstrate the poor impact protection of instrument panels in light aircraft. Most of the paper is taken up with graphic descriptions and photographs of fatal and serious injuries. Pertinent to the present study, it describes an accident involving six young male occupants whose aircraft impacted the ground in a flat attitude after hooking a power line. "None of the six occupants exhibit any sign of external injury. Autopsy studies attributed death to severe trauma to internal organs (brain, heart, liver, spleen, etc) from severe vertical impact forces".

52 White & Panjabi, 1978

Clinical Biomechanics of the Spine

This is a learned, academic work of 515 pages, plus author and subject index. The opening chapters are relevant to this study. Chapter 1 – "Physical Properties and Functional Biomechanics of the Spine" is pertinent in its discussion of the biomechanical properties of the intervertebral disc. Fig 1-15 Vertebral compression strength – from C3 to L5, shows clearly the increase in strength of the vertebrae from the cervical (neck) vertebra C3 to lumbar vertebra L4. At page 26 the statement is made "In general, the vertebrae decrease in strength with age, especially beyond 40 years". Chapter 2 "Kinematics of the Spine" deals extensively with flexion, lateral bending and axial rotation of the spinal column. Citing other researchers, White & Panjabi state "Their results showed an increase in spinal mobility between the decades of 15 to 24 and 25 to 34, followed by a progressive decrease with advancing age".

The researchers also deal with forms of vertebral damage other than axial compression. They point out that many spinal injuries involve edge-damage to the vertebrae. This occurs when the spinal column is not in "optimum" alignment at the instant of impact, i.e. at flexion or lateral bending.

53 Yoganandan N et al, 1990

Injury Biomechanics of the Human Cervical Column

This study relates to motor vehicle accidents, athletic-related events, falls, diving and American football injuries. Tests were carried out on six cadaver cervical columns (it should be noted that, although sports injuries are more likely with younger subjects, the age range of the cadaver subjects was 62 to 77 years). The head-neck complex was submitted to compression – flexion force vectors with a maximum axial input of 4.5kN and maximum piston deflection 3.2cm, hence it is seen that this study is concerned with damage to the cervical column from 'crown-of-the-head' impact. Dynamic fractures of the cervical vertebrae occurred in the range 1.1kN to 3.04kN at approximately 5 to 6ms after impact. The researchers accept that the age range was rather high and that the data require weighting to represent a younger population. Many difficulties had to be overcome to assure alignment of the cervical specimens and the authors stress that neck-muscle structure of live subjects would be a variable.

10 ANNEX 'D'

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