Spinal Injury in Underbody Blast

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Declaration of Originality

I declare that this work is my own, completed with appropriate support from my supervisors. Where other authors' work has been used, it is appropriately referenced.

Copyright Declaration

Acknowledgements

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Abstract

The Improvised Explosive Device (IED) is common in insurgent conflict as such devices are cheap, available, and devastating. Recent literature shows that victims of these devices often have complex injuries including spinal fractures. However, spinal injuries are not described in detail, so the mechanism and effects of these injuries are not well understood.

This thesis reviews the literature with respect to spinal injuries in blast and compares it to UK military victims of IED attacks with spinal fractures. In the UK population, the majority of spinal fractures are thoracolumbar and are associated with multiple other injuries.

This thesis shows that, based on the patterns of injury in UK blast victims, most fractures are caused by a combination of axial loads and flexion, with the apex of the thoracic spine and the thoracolumbar junction most affected by flexion.

Military vehicles incorporate features intended to reduce the effect of blast on their occupants, and a standardised test has been established to evaluate such designs. However, the simple model of the spine used for these tests lacks validity. Understanding the behaviour of the spine in blast incidents will support development of an improved injury prediction model for future vehicle design. In this thesis an *in vitro* study develops a model to understand the role of posture in shaping fracture patterns when the spine is loaded at the rates seen in blast, and supports the mechanistic propositions this thesis makes about the behaviour of the spine in underbody blast.

The clinical outcome and functional effect of blast related spinal fractures is unknown. In a short case series, this thesis suggests that spinal fractures lead to significant pain but the effect of spinal injuries on function is unclear as these victims also have other severe injuries.

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Table of Abbreviations

Fracture Patterns

CF	Compression-Flexion
VC	Vertical Compression
DF	Distraction-Flexion
CE	Compression-Extension
DE	Distraction-Extension
LF	Lateral Flexion
WC	Wedge compression
SB	Stable burst
UB	Unstable burst
TR	Translation
FD	Flexion-Distraction

Abbreviations

ATD	Anthropometric Test Device
CI	Confidence Interval
CT	Computed Tomography
DRI	Dynamic Response Index
EFP	Explosively Formed Penetrator
HANS	Head and Neck Support
IED	Improvised Explosive Device
ISS	Injury Severity Score
JTTR	Joint Theatre Trauma Registry
MDC	Miniature Detonating Cord
NATO	North Atlantic Treaty Organisation
NISS	New Injury Severity Score
PMMA	Polymethyl methacrylate
PRISMA	Preferred Reporting Items for Systematic Reviews and Meta-Analyses
REC	Research Ethics Committee
RR	Relative Risk
SIC	Spinal injury criterion
TRISS	Trauma and Injury Severity Score
USS	United States Ship

1. Introduction

1.1 Scope

This thesis is concerned with the analysis of spinal injuries caused by blast, in particular by explosions beneath vehicles, and with understanding the mechanism of those injuries in order to support injury prediction. The development of mitigation strategies depends on there being a reliable test for changes in vehicle design that will reliably predict whether modifications will result in a reduced risk of injury. The current model is based on a simple system that was developed for testing aircraft ejection seats; the injuries encountered in ejection from aircraft will be reviewed and compared with blast injuries and the injury prediction model reviewed.

1.2 General Introduction

In conventional warfare, mobile armies may be impeded by anti-vehicle and anti-personnel mines laid in order to hinder an invading force. There is a huge variety of such devices available; they may be laid underground, amongst foliage, or dropped from the air and they might be tiny devices designed to wound soldiers or large ones intended to destroy tanks. In insurgent warfare, non-government actors employ similar tactics, using a mix of legacy military weapons and home-made devices to lay Improvised Explosive Devices (IED) to attack their enemies and achieve their goals.

Improvised Explosive Devices are explosive weapons made from a variety of sources, including home-made explosives and military munitions, which can be detonated by local triggers or remote control. They have been a common feature of insurgency in Iraq and Afghanistan, in contrast with previous conflicts where insurgents tended to employ ambush tactics [138, 150, 151] and are likely to continue to be seen in future conflicts and terrorist campaigns. Victims of IED attacks on vehicles have experienced severe high energy limb trauma, which has been the focus of recent research [19, 150, 151].

Spinal injuries in warfare were first reported by the Egyptians around 4000 BC [69]. At that stage, a spinal injury was felt to be a fatal injury and its treatment futile. Although in recent years there have been advances in the management of spinal injuries, little attention has been paid to spinal injuries in warfare. Recent data, however, suggest that spinal injury has become more common, partly because blast victims frequently have spinal fractures amongst multiple injuries (Figure 1) [57, 164, 165, 167-169].

Most papers describing the spinal injury patterns seen in blast lack anatomical and mechanistic detail and so do not enhance our understanding of these injuries [18, 19, 25-27, 141, 145, 146, 163-170]. In addition, there are no recent reported outcome data for victims of spinal injury caused by blast; the most recent paper describing spine related disability in the military population does not separate blast from other causes, nor fractures from muscular back pain [156].

Vehicle designers began to develop features to reduce the effect of anti-vehicle mines and blast as soon as they started developing armoured vehicles [152, 177]. Early attempts to develop protection were primitive, consisting of heavier armour plate aiming to control the interaction between an exploding device and the vehicle, thus reducing the force experienced by the vehicle occupants; these evolved during the Rhodesian conflict [152, 177]. Sophisticated means are now being developed, including energy-absorbing seats to try to reduce the risk of spinal injury [9]. However, the industry relies on a simple test using a model human surrogate dummy to evaluate these developments, and it is not known whether this test is valid. The model does not replicate the behaviour of a real human being, and the injury criterion used to identify whether a spinal fracture is likely is not reliable. An improved test may therefore help further improve protection for vehicle occupants.

Improving the standard test for a mine-resistant vehicle would depend on understanding how the spine behaves when it is subject to blast. At present, this knowledge is lacking, although several authors have published works focussing on specific parts of the spine in specific conditions [63, 170, 179-181, 194-197]. This thesis evaluates the patterns and distribution of injury in the spine in UK victims of blast to start to develop a hypothesis for the mechanism of those injuries. It also demonstrates the mechanism of the most common of those injuries in the laboratory.

There is very little data published regarding the outcome and functional deficit in spinal injury from warfare. A short series is presented in this thesis to start to remedy this deficiency.

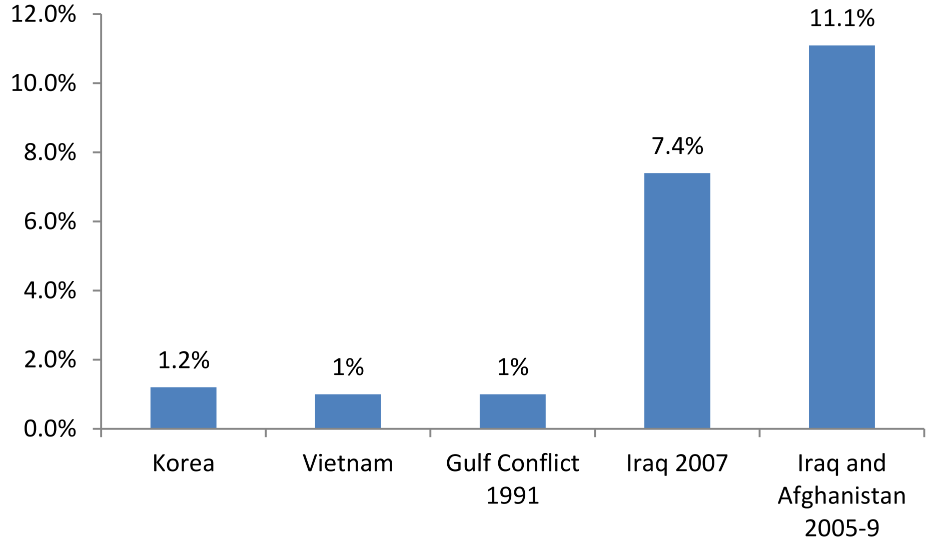

Figure 1: The increasing incidence of spinal injury in US conflicts in key historical publications. Spinal injury received little attention before the Korean War. [26, 164, 169]

1.3 Aims

This thesis aims to improve understanding of the mechanism and effect on victims of spinal injury in blast by reviewing the published literature, identifying injury patterns in blast victims, and re-creating those injuries in an *in vitro* model. Specific objectives include:

- To review the literature relating to the distribution and mechanism of spinal injury in warfare;
- To identify the mechanism of the most common spinal injuries in blast;
- To assess the validity of the standard test for spinal injury risk in blast;
- To identify the clinical and functional outcomes in blast-related spinal fractures.

1.4 Overview

Chapter 2 gives a more details overview of blast injury and the methods used to predict and prevent it. Chapter 3 describes the anatomy and biomechanics of the spine with respect to injury patterns, and then reviews the classification systems used for vertebral fractures. Chapter 4 reviews the published literature describing spinal injury in warfare and Chapter 5 compares those patterns with those seen in UK victims of blast. Chapter 6 considers the current injury prediction model, which is based on aircraft ejection seats, and assesses whether it is fit for its purpose. Chapter 7 describes an experimental study to investigate the mechanism of the most common spinal injury in blast. Chapter 8 attempts to identify the functional effect of blast-related spinal fractures on its victims. Finally, chapter 9 will review the thesis and identify directions for future work.

2. The Physics of Blast Weapons, Vehicle Protection and Injury Prediction

In order to understand blast injury in the spine, it is necessary to discuss the weapons used and the physics of blast, including how they affect a vehicle and its occupants. Therefore, this chapter describes the strategies used to protect vehicle occupants from blast attacks, and the models used to test them.

2.1 Military Weapons

There is a wide variety of explosive weapons in common use. Broadly, these are classified in to "blast" weapons, where the aim is to injure with a blast wave, and "fragmentation" weapons (Figure 2), which are designed to spread fragments with the blast wave to cause injury [66, 98]. Terrorist and insurgent weapons can fall in to either category.

Anti-personnel mines tend to use small explosive charges triggered by the victim, either by a pressure switch on the device or a remote trigger such as a tripwire. They are often designed as fragmentation weapons [46]. Anti-tank mines are typically much larger, and may be simple blast weapons (Figure 3). Typically, anti-tank mines require a much higher pressure to trigger them, so they do not represent a threat to dismounted personnel unless modified. One example of a specific anti-vehicle weapon is the "explosively formed penetrator" (EFP), which was commonly used by insurgents in Iraq (Figure 4). These devices use a shaped charge to melt their casing and propel a high velocity stream of partly molten metal through an armoured vehicle. These weapons often cause devastating injuries to those in the path of the penetrator, but relatively little injury to other vehicle occupants [150].

Figure 2: Anti-personnel fragmentation mine. Note the casing designed to break up on detonation causing secondary blast injury [4].

Figure 3: Anti tank mine, with a large upper surface and a trigger designed not to be fired by a pedestrian. This device is a blast weapon [3].

Figure 4: Explosively formed penetrator. Note the copper dish on top of the device which forms the penetrator [7].

2.2 Improvised Explosive Devices

Insurgents and terrorists use a variety of sources for explosive devices, including homemade explosives and recycled military munitions, with improvised fuses and detonators to suit their new purpose. These are described using the umbrella term “improvised explosive device” (Figure 5). In anti-vehicle applications, weapons are usually triggered by a pressure fuse which may be part of the device or separate and linked by wire. The device explodes when the vehicle tyre triggers the pressure fuse [153]. In Afghanistan, most such weapons were buried [150].

Buried and roadside explosive devices are cheap and readily available in areas of conflict, often making use of legacy military munitions [150]. Devices can be placed at target locations at the operator’s convenience, and then wait for their passive victims. They are therefore eminently suitable weapons for the insurgent, or for armies fighting with limited manpower and financial resources [152].

In anti-vehicle applications, weapons are usually triggered by a pressure fuse which may be part of the device or separate and linked by wire. The device explodes when the vehicle tyre triggers the pressure fuse [153]. In Afghanistan, most such weapons were buried [150].

Figure 5: Military weapons used as improvised explosive devices recovered in counter-insurgency operations. These are shells and landmines with modified fuses. Public domain US Army image [6].

Buried devices apply a significant vertical force to a vehicle and its occupants. The force imparted to a vehicle passenger by the vehicle being accelerated upwards by the blast is significant, and leads to devastating injuries in the lower limbs and the spine [59, 150, 154]. Much attention has been focussed on mitigating these injuries through vehicle design, with a set of standards mandated for testing the injury risk of vehicle occupants in underbody blast [131]. However, these standards have not been validated in the spine.

2.3 Blast Physics

When a device triggers, the high explosive charge in the device detonates (Figure 6). This reaction propagates a supersonic shockwave with the superheated detonation products in its wake [14, 52]. The local pressure wave can exceed 1.2 million psi [14] and exerts a huge force on the soil in which the device is buried. The cap of soil over the device subsequently ruptures, causing a cone of soil ejecta to be propelled upwards behind the shockwave.

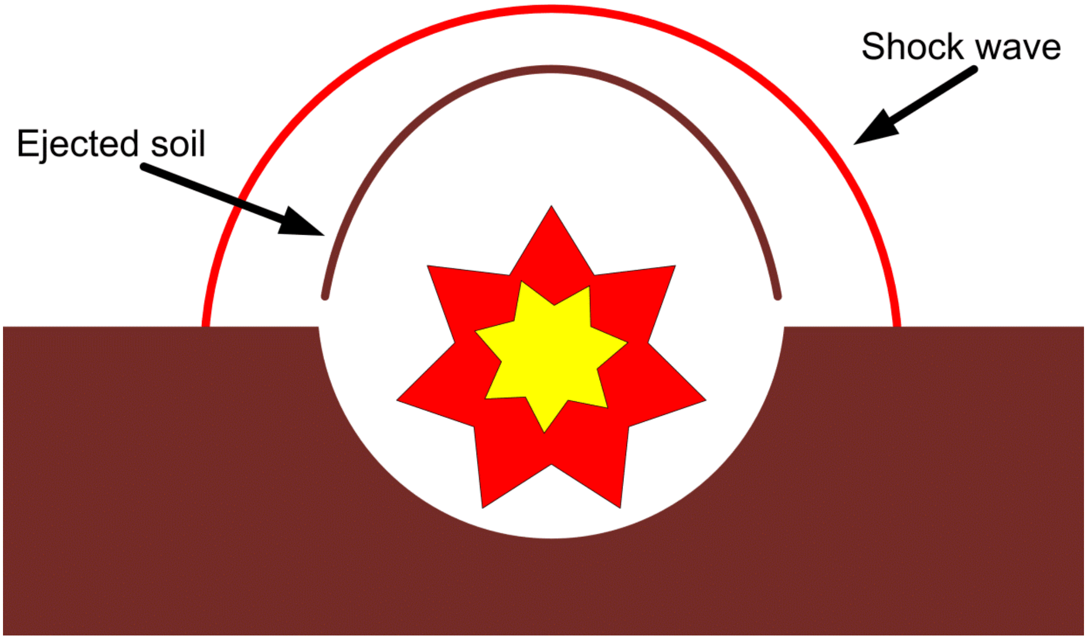

Figure 6: Buried device explosion. The expanding shock wave drives upwards, striking any vehicle above, and is followed by the cap of soil ejecta.

The peak of the shock wave is followed by a short period of underpressure (Figure 7). This underpressure sucks debris with the blast front and contributes to structural and tissue damage [153].

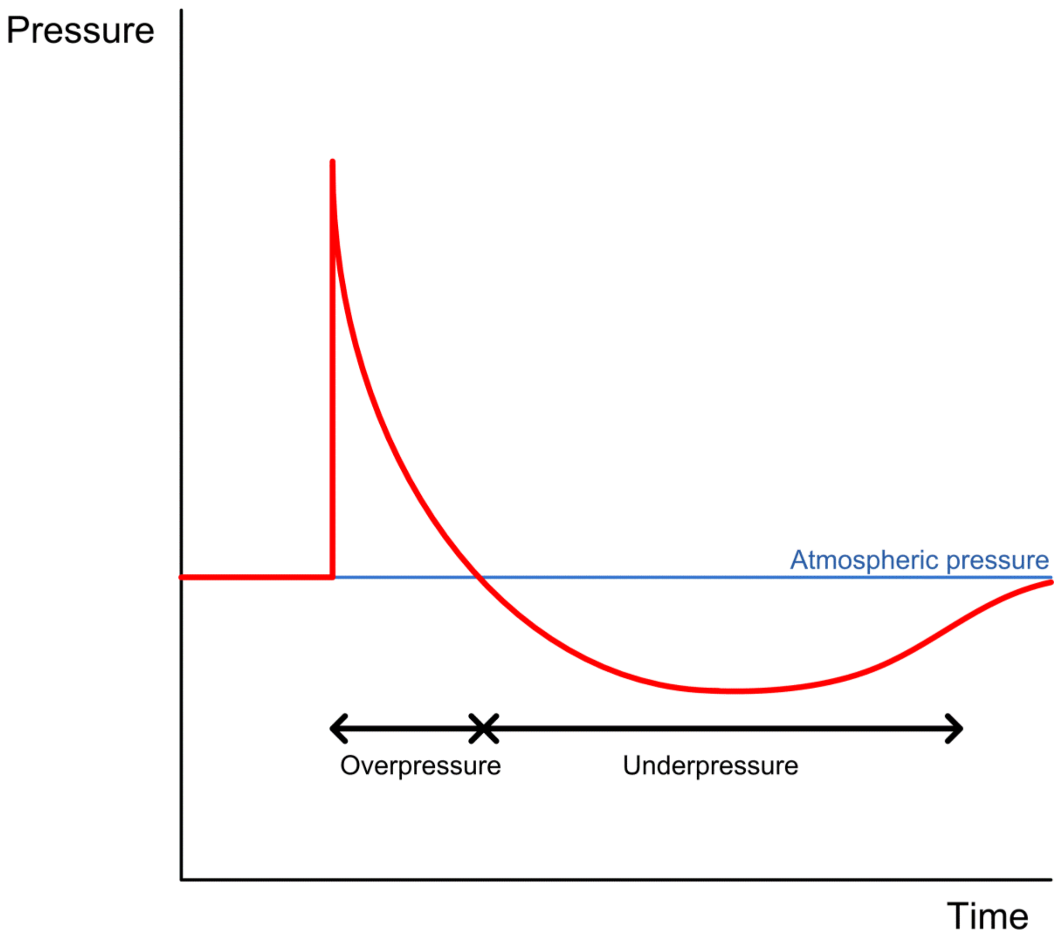

Figure 7: Pressure waveform following blast; this is an idealised curve for a blast in free space showing the almost instantaneous blast overpressure followed by an underpressure wave. Adapted from [64].

In an enclosed space, such as a blast in a building, the peak pressure may be higher than that seen with the same device outdoors [36, 45, 109]. This is because the incident (initial) blast wave reflects off solid surfaces, and the reflected wave joins the incident blast wave with constructive interference (Figure 8), producing a higher pressure than the primary blast wave alone [142]. The front that this forms is known as a "mach stem". This effect is also seen when a device detonates near a wall. Victims of blast in enclosed spaces, or standing near walls, are therefore subject to a high blast pressure, for a longer duration, and may be struck by more than one wave, leading to more severe injuries [36, 83]. The same is true for a blast under a vehicle; because the wave is affected by the ground and the vehicle itself, the actual pressure wave seen by the vehicle can be very different to the idealised waveform shown in Figure 7. However, the way in which the wave and soil mass interacts with such a complex environment is difficult to measure, and therefore difficult to model and predict.

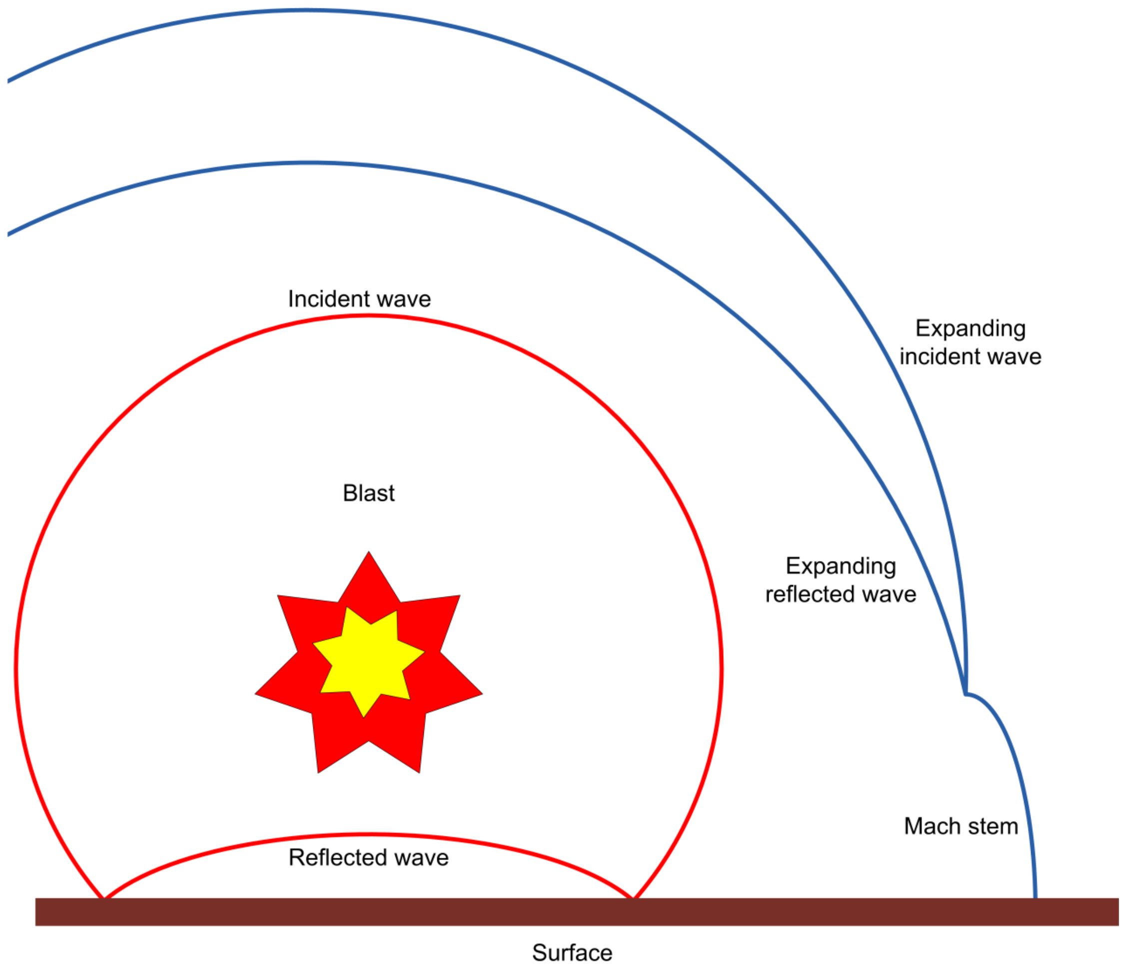

Figure 8: Mach stem effect, showing an idealised free field blast interacting with a flat surface. The expanding incident wave combines with the reflected wave through constructive interference to form the Mach stem. Adapted from [142].

2.4 Blast Injury

There are several types of injury following blast [203]. *Primary blast injury* occurs as the shockwave passes through the victim, depositing energy and damaging tissues at interfaces between tissues of different density. *Secondary blast injury* is caused by fragments propelled by the blast wave striking the victim. *Tertiary blast injury* occurs when the victim is moved by the blast wind and strikes another object or the ground. *Quarternary blast injury* describes other effects of the heat and fragments in blast, such as burns, chemical effects of fragments, and infection.

2.4.1 Underbody Blast Injury

When a vehicle detonates a buried device, energy is transferred by the shockwave and mass of soil ejecta to the vehicle, leading to deformation of the vehicle floor and acceleration of the vehicle upwards (Figure 9). Injuring a victim indirectly through force transmitted by the vehicle was described as "solid blast" during World War Two [15].

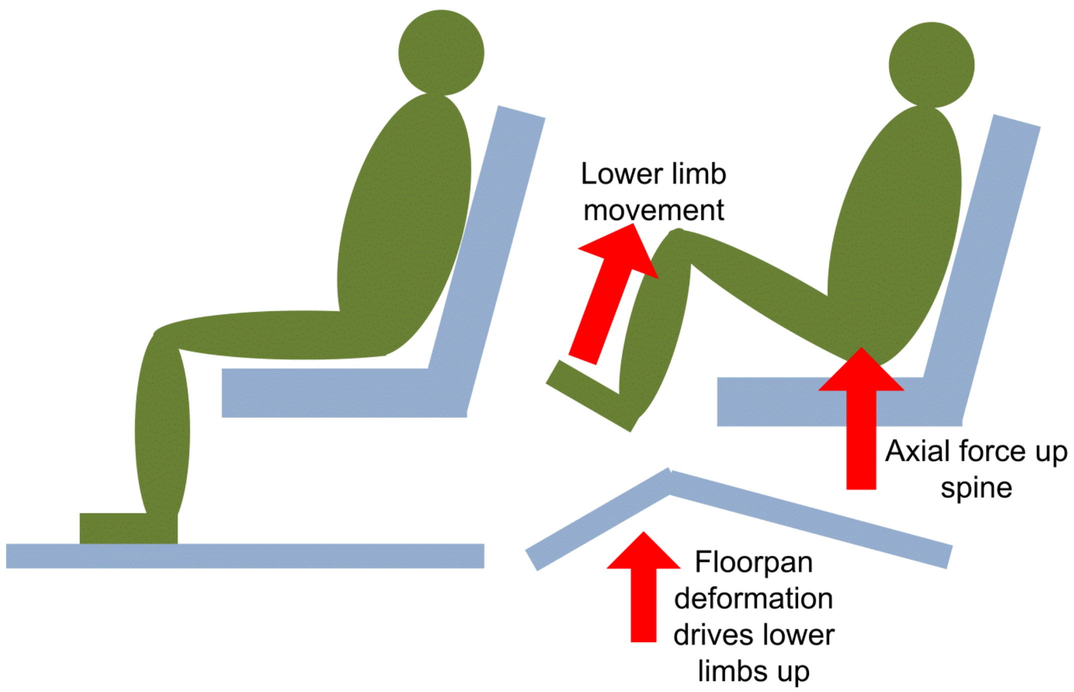

Figure 9: Effect of underbody blast on a seated victim, showing the axial force along the axis of the spine combining with floorpan deformation driving the limbs upwards and rotating the pelvis, probably altering the posture of the spinal column.

Spinal injury in solid blast is not well understood. Part of the force is directed through the seat pan to the pelvis and spinal column, leading to vertical acceleration of the spine and torso [108, 149, 152]. The floorpan deformation may also drive the lower limbs upwards, leading to hip flexion and subsequent rotation of the lumbar spine [108, 149]. There may also be injury as the victim strikes the vehicle roof and injury as loose fragments strike the victim, such as head injury, which may be a significant cause of death in these victims [173]. However, the contribution each of these events makes is unclear, so the overall mechanism of spinal injury is uncertain. This thesis will endeavour to improve this understanding.

2.5 Injury Prevention

Since World War One, armoured vehicles have evolved to counter the developing threats against them. Early tanks used steel plate armour to resist penetration by solid and explosive shells, preventing secondary blast injury (Figure 10). Modern armour includes complex composite materials designed to counter kinetic energy weapons such as armour piercing shells and EFPs, and spaced cage armour to counter high explosive anti-tank weapons.

Figure 10: British Mark 1 tank on the Somme. Note the wire mesh to protect against dropped hand grenades. Copyright expired image [70].

Figure 11: Coyote vehicle in Afghanistan. Note the V shaped hull, high ground clearance, and blast deflectors, all designed to reduce the effect of underbody blast. Public domain image [40].

In recent years, attention has been paid to designing vehicles which aim to mitigate the effect of underbody blast on occupants. One example of a vehicle design feature aimed to reduce the transfer of energy in this interaction is the V-shaped vehicle hull, intended to dissipate the blast wave around the vehicle (Figure 11 and Figure 12) and therefore reduce the amount of energy transferred to the vehicle body, thus reducing tertiary blast injury. This was shown to be effective in the Rhodesia conflict [177]. Other design features that were shown to be effective include increasing vehicle mass, raising the vehicle from the ground, and installing blast protectors over the wheels. Each of these, however, introduces compromises in vehicle design, producing ever larger and heavier vehicles with subsequent difficulty manoeuvring in limited spaces [153].

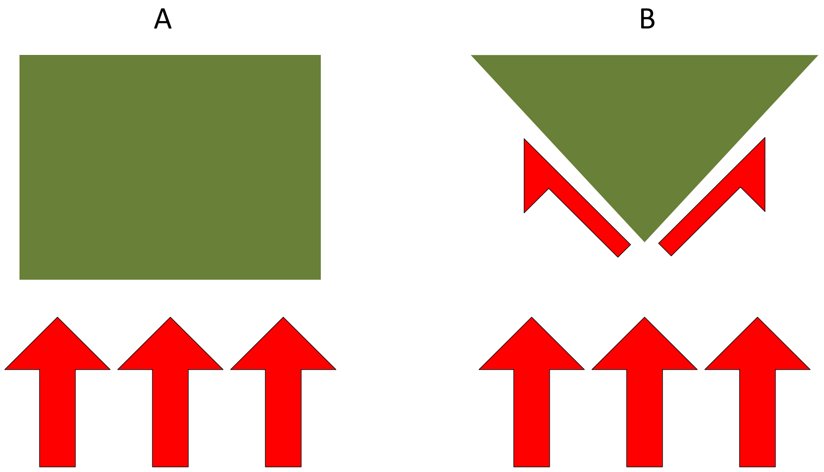

Figure 12: Blast wave interaction with vehicle floor. A: Flat vehicle floor receives all the energy from an underbody blast. B: A V-shaped hull allows energy to dissipate around the v shape. Adapted from Ramasamy [153].

2.5.1 Injury Mitigation in Civilian Environments

In the civilian environment, spinal injury mitigation targets high risk activities such as sports and motor racing with devices that focus on impact protection, aiming to reduce the effect of an impact load on the spine beneath.

Motorcyclists and horse riders both employ back protectors to reduce the risk of spinal injury following a fall. One Italian study has shown that they are effective in reducing injury risk from a fall landing on the device [67], but they cannot protect against axial load impact while still in the saddle.

A more recent development intended to reduce the risk of injury by controlling movement during impact is the Head and Neck Support (HANS) device from HANS Performance Products, New Braunfels, TX, USA. This neck brace acts in conjunction with a harness to prevent head and neck flexion, and is intended to reduce the risk of basal skull and cervical spine fractures during impact [172].

Perhaps the best known example of spinal injury mitigation is the seatbelt. The modern three-point belt was developed for Volvo by Nils Bohlin, originally an ejector seat designer, in the 1950s [28]. They have been shown to significantly reduce the risk of spinal injury during a motor vehicle collision [128]. Seatbelts help to control both load and the range of movement of the spine during impact, thus reducing the risk of injury caused by excessive flexion movement.

2.6 Current Standards for Injury Prevention

The North Atlantic Treaty Organisation (NATO) is the overarching organisation for Western military forces. In order to standardise equipment between nations, NATO sets specifications for a variety of equipment including weapons and vehicles. NATO defines standards for testing vehicles to predict the injury burden to the occupants under blast load [131]. These tests are based around the standard Hybrid III anthropometric test device (ATD), shown in Figure 13, which was originally developed for automotive crash tests.

Figure 13: Hybrid III anthropometric test device on a drop tower. The simple cervical spine is clearly visible. Author's photograph.

NATO document AEP-55 [130] defines procedures for testing vehicle resistance to mine threats. Specific tests are defined for tanks and light armoured vehicles, stating size and position of explosive charges for whole vehicle tests, and defining the seating type and position for the Hybrid III ATDs used in the test. Each dummy is fitted with load cells in the spine, and an accelerometer in the pelvis, to measure the forces applied to the dummy spine during loading. Load cells elsewhere are used to predict injury in the limbs. The peak measurements in these sensors are used to predict the risk of critical injuries including tibial fractures, thoracolumbar spine fractures, neck injury, and injury to the internal organs. The correlation between injury and measurements from such sensors has been developed from biomechanical research, much of which is from the automotive industry.

The standards for spinal injury are based on data from ejection seat tests [131] which was developed for much lower loading rates than are seen in blast; recent evidence suggests that the model is not appropriate for blast injury and there is therefore a need to develop a more biofidelic model of underbody blast injury in the spine [88, 131, 200].

2.7 Injury Prediction Models

The risk posed by vertical acceleration of the spinal column was first considered in 1944 [182]. The first tests were conducted using horizontal deceleration sleds to approximate a 12g deceleration from 65 feet per second. Subsequent tests were conducted by Martin Baker using live subjects on upright acceleration towers [144].

Henzel [82] reviewed the bioengineering literature with regard to ejection seat injury in 1967. The data available at the time supporting the load to failure of the vertebra was mostly produced by quasistatic (extremely low loading rate) loading experiments. The paper introduced several injury risk prediction models, of which the most important is the Dynamic Response Index (DRI), later adopted as the standard for predicting spinal injury risk in underbody blast [131].

2.7.1 Dynamic Response Index

The Dynamic Response Index (DRI) developed by Stech and Payne [178] was adopted by NATO in 1973 for ejector seat tests, and later used for underbody blast testing [11]. This attempts to represent the maximum dynamic compression of the thoracolumbar vertebral column during axial acceleration in order to estimate the risk of spinal injury. The model uses a single mass-spring damper system to predict the gross response of the spine following a short duration pulse load (Figure 14) [31]. The model is a one-degree of freedom model considering only axial acceleration and has been developed using data from various sources including cadaveric studies. The schematic model used in DRI is shown in Figure 15.

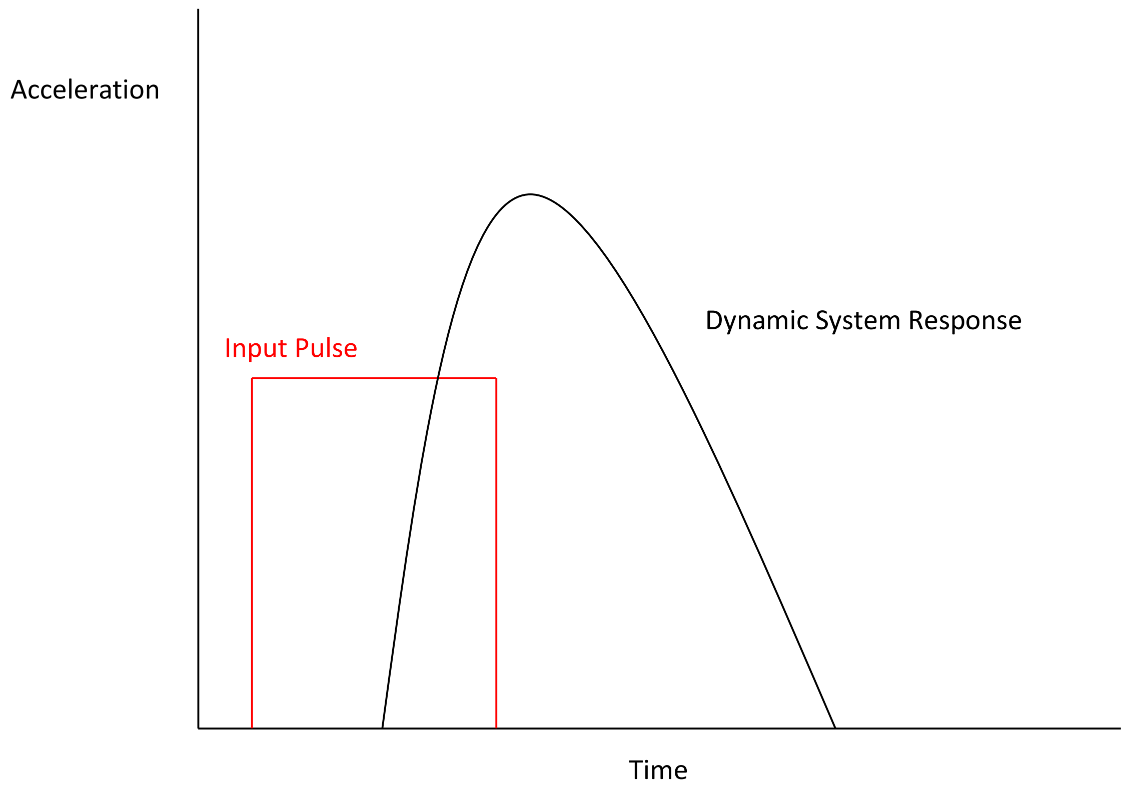

Figure 14: Conceptual input - response curve for the spine used in DRI. This hypothetical graph shows a sudden, short impact causing a delayed response from the loaded system, much as the series of springs and dampers that make up a modelled spine might behave following blast. [178].

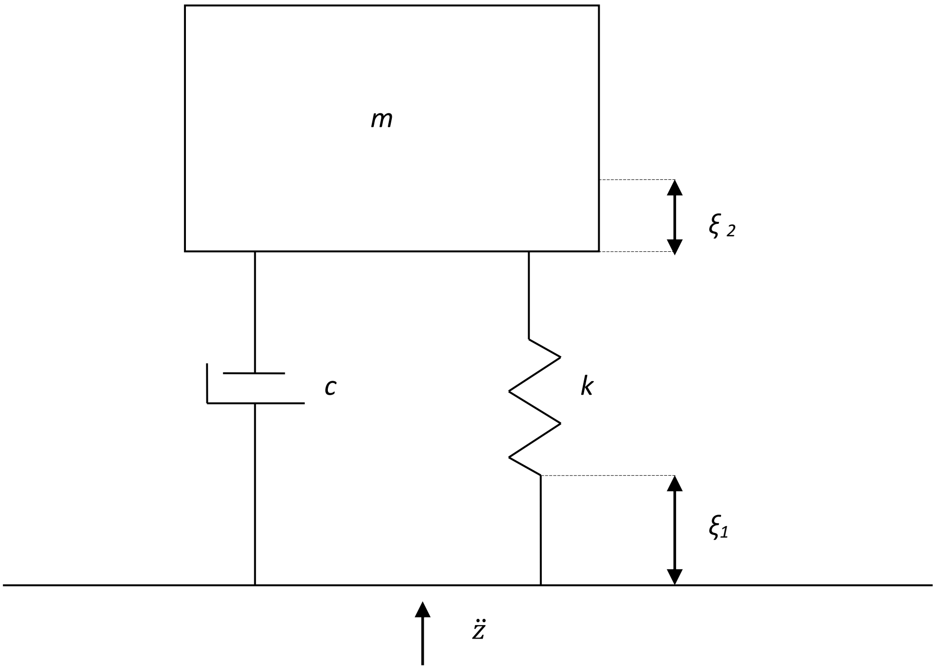

Figure 15: Dynamic Response Index model [131] c = the damping coefficient of the system, k = the spring coefficient, ξ are the displacements of the floorpan and the body mass caused by acceleration of the pelvis, $\ddot{z}$.

The equation of motion for the DRI is:

$$\ddot{z}(t) = \ddot{\delta} + 2.\zeta.\omega_n.\dot{\delta} + \omega_n^2.\delta$$

DRI is calculated as [131]

$$DRI = \frac{\omega_n^2.\delta_{max}}{g}$$

Where

δ is the relative displacement of the system, $\delta = \xi_1 - \xi_2$

ζ is the damping coefficient $\zeta = \frac{c}{2.m.\omega_n}$

ω_n is the natural frequency with $\omega_n = \sqrt{\frac{k}{m}}$

Stech and Payne specified the values of ω_n as 0.224 Hz and ζ as 52.9 rad/s for a representative US Air Force pilot aged 27.9 years. These values were derived from repeated experiments using cadavers of different ages and in different postures [178], which sought to identify the resonant frequency of the spine. A DRI value of 21.3 was predicted to give a fracture rate of 50% based on cadaver tests. The risk of injury was produced as an approximated curve by Stech and Payne [178] and is shown at Figure 16.

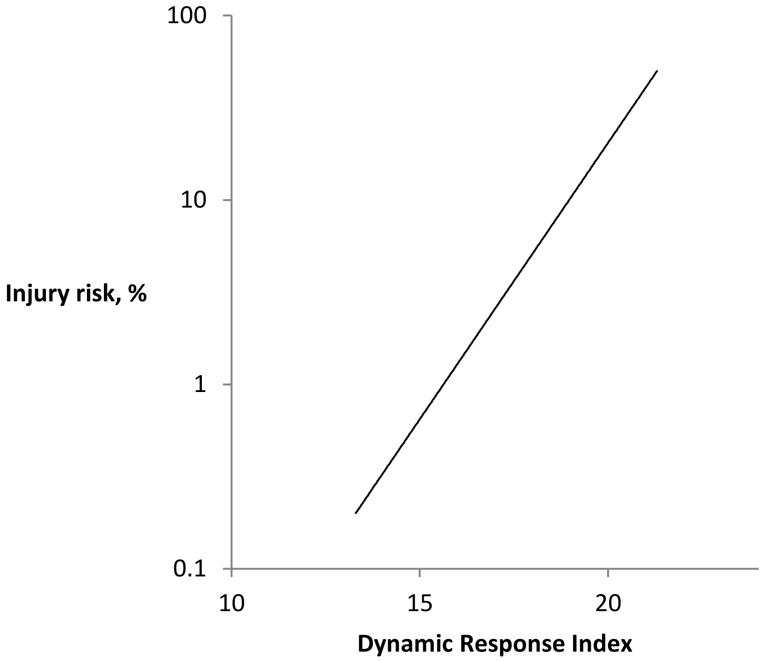

Figure 16: Simple injury risk curve using DRI suggesting that a DRI value of 21.3 is associated with a 50% probability of injury [31, 178].

The Hybrid III anthropometric test device – “dummy” - was designed to evaluate injury risk in the automotive industry. Several derivatives of the dummy exist with modifications aimed at blast and ejection seat testing. Specifically, the automotive dummy is fitted with a curved lumbar spine; there is an aerospace crash test derivative with a reduced lumbar lordosis [155]. However, the spinal column in the dummy was not designed to reproduce the behaviour of the human spine under axial loads. The Hybrid III lumbar load cell does not measure the total force imparted by the torso to the lumbar spine during an axial deceleration test [155]. Since limited data are available on the behaviour of the spinal column under blast acceleration, a high-fidelity spinal surrogate has not yet been developed. The dummy spinal column cannot exhibit identical resonance behaviour to the human spinal column and therefore its interaction with a seat system and torso mass is unlikely to represent the behaviour of a living victim. Further, DRI assumes that injury is caused solely by axial acceleration; if, as suggested above, injury is affected by the lower limbs moving and causing spinal flexion, the DRI has no way to account for this.

Several attempts have been made to improve on the DRI model with respect to blast injury but none has yet been found to be satisfactory. The Spine Injury Criterion (SIC) was established using a Hybrid III dummy in drop tower tests with a pelvic accelerometer and lumbar spine load cell [55, 131]. This system correlated pelvic acceleration with lumbar spine compression force. However, injury prediction was still based on the data used in DRI. Chandler correlated peak compression force at the ATD spine to injury risk [39]. Neither of these models has been shown to be better than DRI.

A more recent high-fidelity model has been proposed by Zhang *et al.* [201]. This model used an idealised, symmetrical computer model of a lumbar spine to predict injury in underbody blast but the model is somewhat simple, with limited contribution from surrounding anatomy, and the material properties of the components of the model are based on quasistatic loading models which may not apply at the loading rates seen in blast. Like SIC, this model has not been shown to improve on DRI.

2.8 Summary

This chapter has discussed blast weapons and the physics of blast, as well as the models of blast on victims. Blast attacks on individuals and vehicles produce injury from the effect of blast wave, from fragments, and from moving the victim. Blast produces a large spectrum of injuries throughout the body; while recent research has focussed on the lower limb, this thesis aims to develop understanding of the effect of blast on the spine.

When a vehicle is struck by underbody blast, its passengers are at risk of spinal injury. The mechanism of these injuries is unclear, but may involve axial force along the spinal column along with limb movement as a result of the vehicle floor plan deforming. In order to attempt to reduce the risk of these injuries, vehicles incorporate specific design features that are tested according to standard protocols using accepted injury prediction models. There are several such models in use for estimating the risk of spinal injury after underbody blast attack, of which DRI is the most widely accepted and the model mandated for use by NATO.

The next chapter will review the anatomy and biomechanics of the spine to allow better understanding of the injury patterns discussed in this thesis.

3. Classifications and Mechanisms of Spinal Injury

In order to understand the patterns of injury in the spine, and to relate them to the mechanism of injury, it is necessary to review the anatomy and the biomechanics of the spinal column. This chapter gives a basic overview of the anatomy and behaviour of the spine. The main classification systems used in describing spinal fractures are then discussed.

3.1 Spinal Column Anatomy

The spinal column consists of seven cervical, twelve thoracic, and five lumbar vertebrae (Figure 17). The first cervical vertebra articulates with the base of the skull and the fifth lumbar vertebra articulates with the sacrum, which is formed of fused vertebrae but is not part of the mobile spinal column. This thesis will not consider the sacrum as it is functionally part of the pelvis rather than the spine.

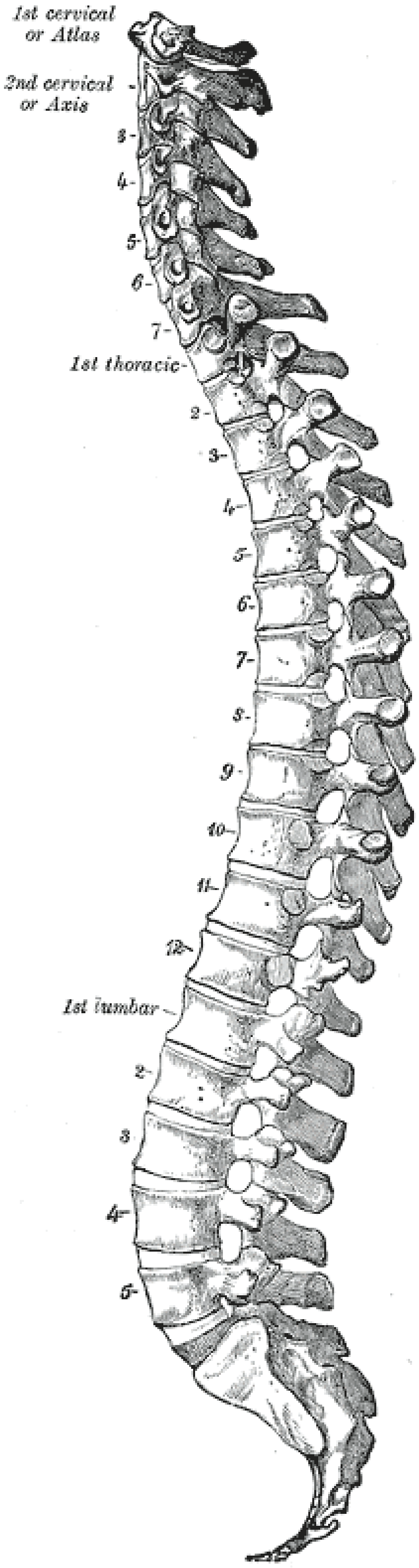

Figure 17: The spinal column showing the lumbar and cervical lordosis and the thoracic kyphosis. [1]

3.1.1 Vertebrae

There are seven cervical vertebrae. The first and second vertebrae are specialised and have different morphology than the lower vertebrae. The first cervical vertebra is the atlas (Figure 18) which articulates with the occipital condyles of the skull above at the atlantooccipital joint. This joint is responsible for a significant proportion of head flexion and extension; there is limited lateral and rotational freedom at this level.

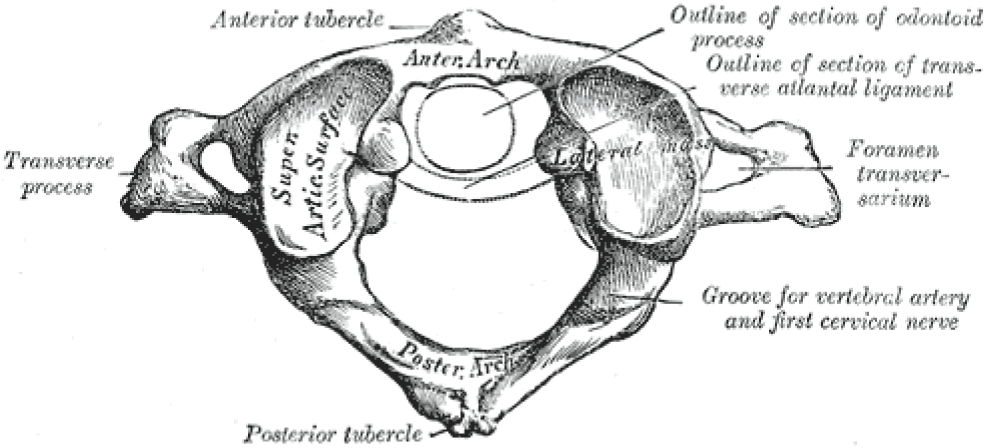

Figure 18: C1 vertebra, the atlas, from above [73].

Below the atlas is the C2 vertebra, the axis (Figure 19). The two articulate through the unique atlantoaxial joint between the peg (dens) of the axis and the atlas. This joint is responsible for nearly all rotation in the neck [174].

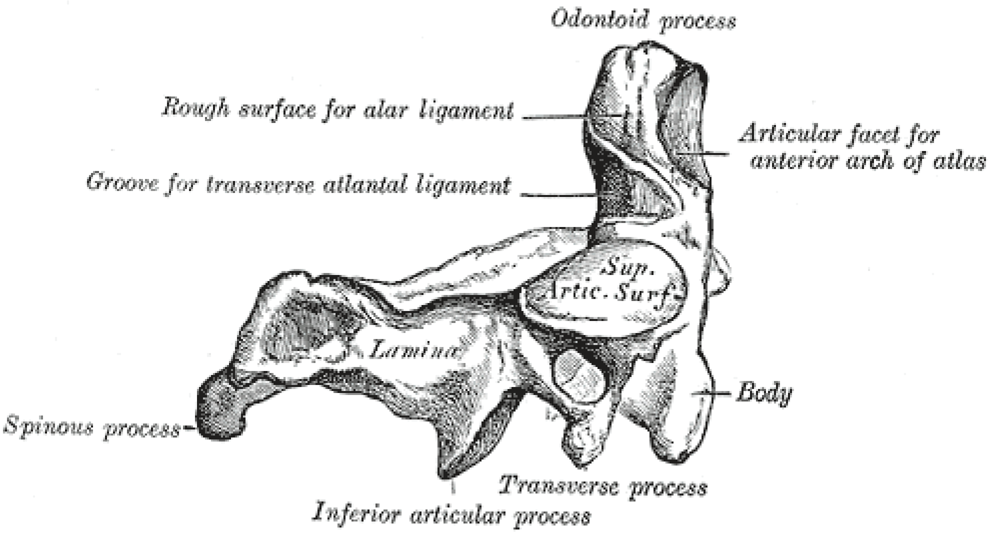

Figure 19: Axis (C2 vertebra) showing the dens about which the C1 vertebra rotates [73].

The remaining vertebrae in the cervical spine are described as “subaxial” [92]. A typical subaxial cervical vertebra is shown in Figure 20. Each vertebra articulates with its neighbours through the body, joined by the intervertebral disc, anteriorly. Posteriorly each vertebra has a superior articular surface which articulates with the inferior surface of the vertebra above to form the facet joint.

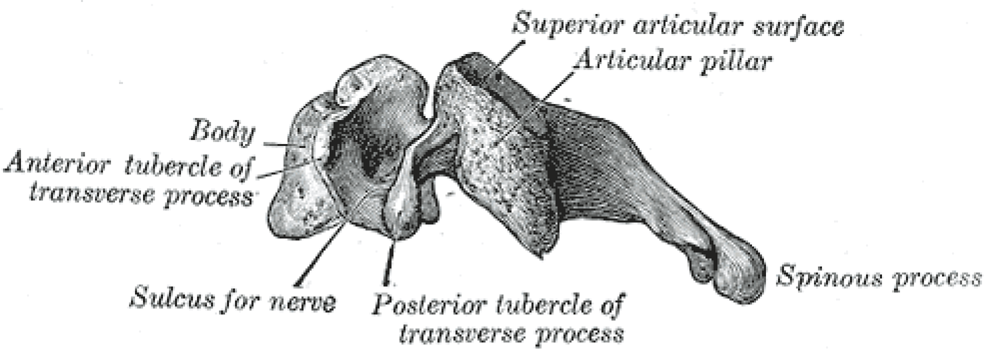

Figure 20: Typical cervical vertebra, lateral view [73]

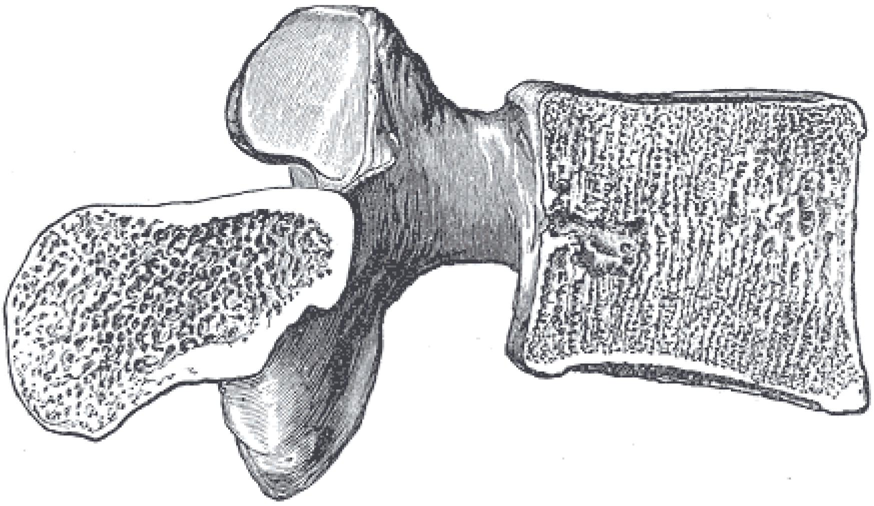

Figure 21: Cross section of vertebra showing trabecular and cortical bone [1]

The vertebra can also be divided in to anterior and posterior elements. The anterior element is the body. This consists of a cortical shell surrounding a trabecular bone core, filled with marrow in younger patients (Figure 21) [127].

The posterior elements, joined by the pedicles to the body, include the spinous process, providing insertion for the strong posterior ligaments, and the transverse processes, providing insertion for the spinal muscles. In the thoracic (Figure 22) and lumbar (Figure 23) vertebrae, the articular processes are attached close to the junction of the pedicle and transverse process. The part of the vertebra between these is the *pars interarticularis*.

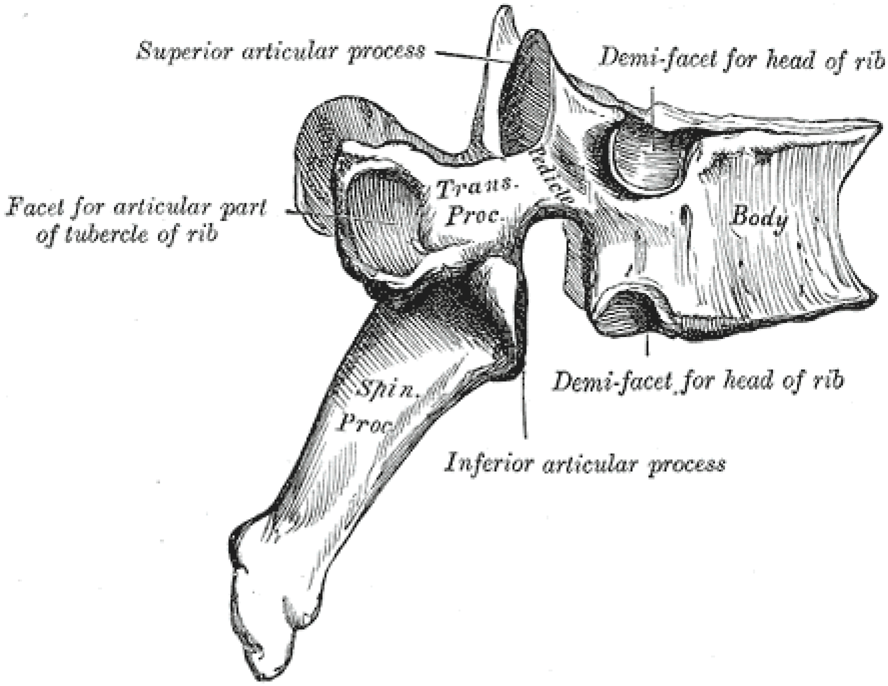

Figure 22: Typical thoracic vertebra shown from the side, showing the rib articulations [73].

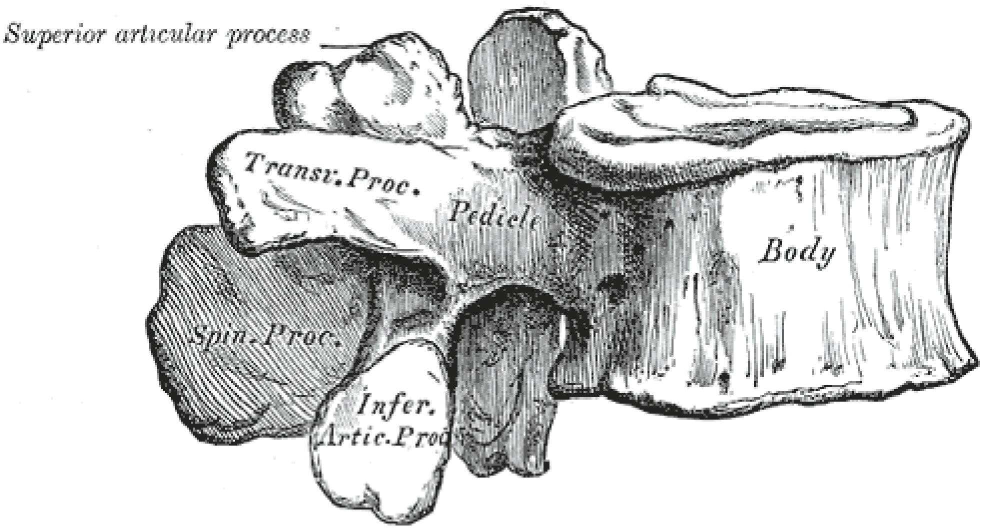

Figure 23: Typical lumbar vertebra showing the larger body and pedicles compared to other vertebrae [73].

The spinal cord runs in the foramen formed by the pedicles and laminae. The cord itself terminates in a conical end known as the *conus medullaris* between the first and second lumbar vertebrae. Thereafter the nerves continue as the *cauda equina.*

The morphology and mechanical properties of vertebrae change along the length of the spinal column. Thoracic vertebrae have an articulation with the adjoining rib. The vertebral body is larger in the lower part of the spine, reflecting the increasing loads borne by lower vertebrae. The facet joint alignment also changes along the vertebral column.

3.1.2 Ligaments

The vertebrae are joined by a series of strong ligaments (Figure 24) which provide stability and resistance to excessive flexion, extension, rotation, and shear. The anterior and posterior longitudinal ligaments adhere to either side of the vertebral bodies. These ligaments are relatively inelastic. The *ligamentum flavum* runs down the posterior part of the spinal canal; it has significant elasticity [190]. The spinous processes are joined by the interspinous and supraspinous ligaments. These last three ligaments are described as the "posterior tension band" and resist excessive spinal flexion [118].

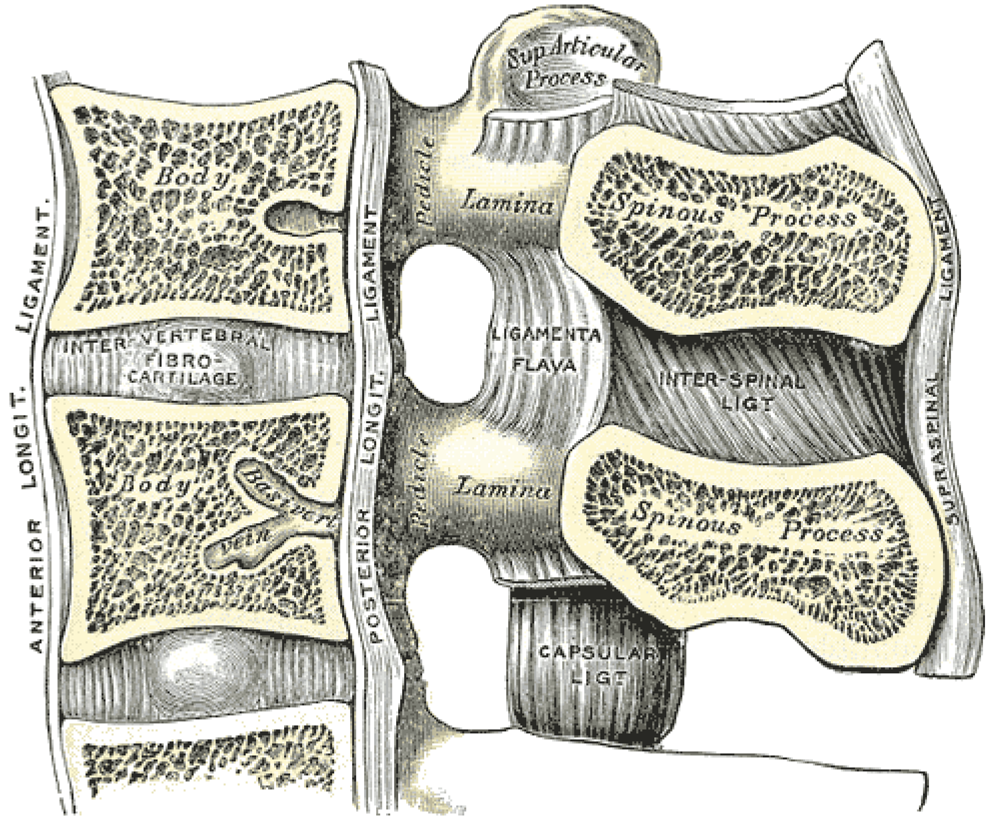

Figure 24: Sagittal section of lumbar spine showing intervertebral disc and ligaments [72]

3.1.3 Intervertebral Disc and the Motion Segment

Anteriorly the vertebral bodies are linked through intervertebral discs (Figure 25). These consist of two parts, the annulus fibrosus, which is constructed of obliquely aligned collagen fibres, and the central nucleus pulposus, a gelatinous mass the outer annulus is arranged in lamellae of collagen fibres and resists the spreading force of the central nucleus pulposus. The disc is attached to the vertebrae peripherally, with the annulus fibres connected to the body. The part of the vertebral body adjacent to each disc, forms the endplate with strong cortical bone. During spinal motion the disc demonstrates significant flexibility and is under compression in load bearing activity [8] It is also critical for shock absorption of the impact loads experienced during normal motion.

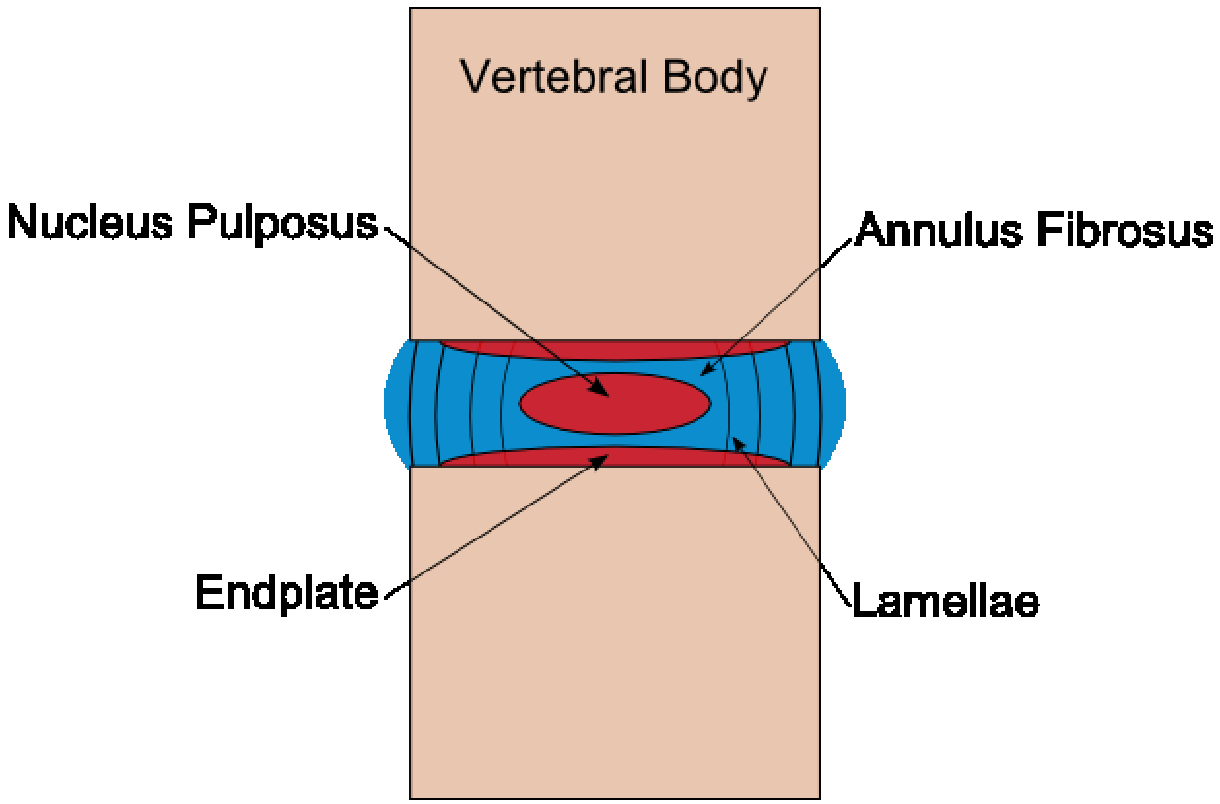

Figure 25: Simplified schematic diagram of intervertebral disc

Two adjacent vertebrae, joined by a disc, are defined as a “motion segment” and often referred to as a “functional spinal unit” [8]. During spinal movement, the disc allows movement between vertebral bodies and the facet joints posteriorly slide relative to each other.

3.2 Mechanisms of Fracture

Attempts were made to assess the biomechanical limitations of the vertebral column as early as 1880, when Messerer [123] noted that the breaking strain of cadaveric vertebrae was higher in the lumbar than cervical spine. Several clinical and bioengineering authors have since examined the failure behaviour of the vertebral column, and of individual vertebrae. Much of the literature surrounding spinal injuries is based on sporting injuries or those sustained during a fall from height, with an axial load to the spinal column [25, 110].

Under compressive loads, the vertebral body transmits most force [190]. Most load is transmitted by the compact cortical bone, especially in the centre of the vertebral body [34]. The cancellous centre of the vertebral body is able to deform significantly before failure [79, 115], whereas cortical bone deforms less than 2% before failure. It is not clear, however, which order the components of the vertebral body fail in during real-world injury, nor if the cancellous bone actually deforms in real loading situations. The cancellous core also contains bone marrow, which moves through trabeculae as they collapse during bone structural failure, providing a hydraulic cushion which at high rates of loading resists dynamic peak loads [80, 100, 190].

There is a significant change in the failure behaviour of bone as the strain rate increases. Studies on bovine cortical bone [199] showed an increasing elastic modulus at loading rates between 1000 and 2000 mms^{-1} with a tendency to brittle failure at higher loading rates. Strain at failure was highest at 1500 mms^{-1} with a reduction in toughness at higher loading rates. Human bone experiments [78] have shown similar results with the Young's modulus increasing as the strain rate increased under compression. Under tension, the modulus decreases as the strain rate passes 1 s^{-1}. Cancellous bone deforms under compression [10, 115], but tends to undergo brittle failure under shear or tension [10]. Fracture comminution has been shown to be proportional to the energy absorbed prior to failure [10, 161] in quasistatic loading experiments. Low strain rates produce linear fractures; higher rates are associated with comminution.

Kazarian and Graves [100] studied the failure of isolated vertebral bodies under high strain rate loading. Each vertebra was loaded at deformation rates of up to $0.9ms^{-1}$ until 50% height loss was reached. It was shown that stiffness and breaking strength increased with a non-linear relationship as strain rate increased. This paper focussed on the behaviour of the fluid within the trabecular meshwork and the influence of its flow path as the vertebra is compressed.

A more recent study of the high strain rate behaviour of failing cancellous bone [84] used a split Hopkinson pressure bar to investigate failure at strain rates up to 1000 s^{-1}. In this study, again, compressive strength increased with loading rate.

Three papers giving breaking strain of individual vertebrae are summarised at Table 1. Each of these papers shows quasistatic loading behaviour where the vertebra is loaded slowly. There is a wide variation of load to failure between these studies, relating both to methodology and age of the specimens used.

	Failure load, N		
Vertebra	**Yamada [192]**	**Gozulov [71]**	**Messerer [123]**
C1	4092	7829	
C2	4092	4991	
C3	4092	3954	1468
C4	4092	3995	2162
C5	4092	4435	1664
C6	4092	5511	1664
C7	4092	4542	1957
T1	3620	4648	1957
T2	3620	4266	1859
T3	3620	4568	2055
T4	3620	5107	2055
T5	4216	5391	2153
T6	4216	6058	2447
T7	4216	6663	2447
T8	4216	8065	3132
T9	6303	8220	3523
T10	6303	8416	3914
T11	6303	8972	3670
T12	6303	10315	3914
L1	7143	10658	4159
L2	7143	11499	3425
L3	7143	11743	3914
L4	7143	11743	4159
L5	7143	12606	1468

Table 1: Breaking force for individual vertebrae under quasistatic load from three historical papers [100]. All expressed in N, converted from imperial units where appropriate.

The vertebral endplate has been shown to fail before the disc under both quasisatic and high-rate loading [143, 190]. In young patients the load to failure of the endplate under quasistatic loading was approximately 4000 N; this increased to 13500 N if the load was applied over 0.006 seconds. With an intact disc, the centre of the endplate fails first. In degenerate discs, the nucleus is less able to exert fluid pressure on the endplate which therefore fails peripherally. This appears also to be the case at higher loading rates.

The facet joints are important stabilisers of the motion segment [190] and limit motion of the posterior parts of the spine during flexion [148]. In static conditions the facets transmit 18% of the compressive load though each lumbar spinal level [129]. Dynamic studies have shown that the relationship between facet and disc load is complex [104] and depends on the spinal posture; in significant flexion the facet joints are completely unloaded and the posterior ligaments are in tension. The posture of the spine at the moment of failure is therefore relevant to the pattern of injury.

Roaf [157, 158] published two papers describing the mechanisms of fracture in the spine in which some of these mechanisms are considered. These papers described the endplates failing with quasistatic loading and consider the effect of posture on the failure pattern of the vertebrae. This concept was reviewed by Hoshikawa *et al.* [85] in a paper which discussed, under quasistatic axial loading, the change in fracture pattern as the flexion angle of the spine was increased. It was shown that with sufficient flexion the anterior part of the vertebra fails in compression, but the posterior elements fail in tension.

Adams et al. [8] also reviewed the effect of posture on the compressive strength of the lumbar spine. This study used quasistatic loading of lumbar spine motion segments to assess the distribution of load sharing between the disc and facet joints. Flexion beyond 75% of the segment's normal range led to high tensile forces in the posterior ligaments and an increase in the disc pressure. Significant extension led to facet joint fracture at loads as low as 500 N as they became load bearing. The paper concluded overall that moderate flexion is ideal for resisting axial load in the lumbar spine. There was a wide variation in compressive strength of L4 and L5 in these specimens, ranging from 2704 to 10064 N. It was also noted that the rate-dependent properties of the disc and motion segment in post-mortem specimens in this study may differ from those in vivo [101, 102] but the effect is unclear. Flexion in the lumbar spine has been shown to cause shear at the disc-endplate interface [107]. This is resisted by the orientation of the facet joints.

One review of the biomechanics of cervical spine fracture considered the effect of both posture and loading rate. Maiman *et al.* [119] loaded the apex of the skull at rates between 0.25 and 152 cms^{-1} on flexed and extended cervical spines. It was noted that flexed spinal units failed at lower loads than extended spines and that lower loading rates lead to failure at lower loads than higher rates. Nonetheless, the highest loading rates used were significantly lower than would be expected in blast injury. Most of the spines that failed in flexion had posterior ligament failure.

Since the posterior elements often fail in tension, it is worth considering the failure strength of the spinous processes under tension loads. Golish *et al.* [68] loaded the spinous process to failure in a static experiment and showed a mean load to failure of 453 N.

Vertebral failure in axial loading is therefore subject to a variety of factors which affect the pattern and risk of fracture. A force aligned with the vertebral neutral axis is likely to produce a compression fracture. Anteriorly directed forces introduce flexion and lead to anterior vertebral failure [179] perhaps combined with posterior fracture or ligamentous injury. The spinal column is likely to move during the application of axial load, so the force vectors causing failure change during loading of the spinal column. The fracture pattern may therefore allow the axis of the causative force for each fracture to be derived. The strain rate dependent behaviour of the vertebra as it fails is less well understood, and this may be important in applying the standard mechanistic classifications of injury to blast victims as these classifications are normally derived from civilian, low energy injuries.

3.2.1 Ligament properties

The spinal ligaments act to restrict the range of intervertebral movement. Ligaments are uniaxial structures which resist tensile forces in the line of their anatomical alignment [190]. Spinal ligaments demonstrate nonlinear mechanical properties with stiffness increasing with the applied strain [183, 184].

The anterior and posterior longitudinal ligaments share similar material properties. The ligamentum flavum is significantly more elastic [190], with a resting tension that produces compression of the intervertebral disc. Along with the supraspinous and interspinous ligaments, the ligamentum flavum is under tension during compressive loading of the spine. The ligaments surrounding the facet joint capsules, especially in the lumbar spine, are placed under tension during axial loading and have been shown to stretch during ejection from aircraft [148].

Testing porcine ligaments to failure [29] has shown similar results, with increasing stiffness and load to failure as strain rate increases. This property has been suggested to help protect the spine against excessive movement or vertebral load during transient impact [184]. It may also be relevant in the causation of avulsion fractures, where bone failure occurs at the insertion of a ligament subject to a high tensile load; this is likely to be the mechanism of most spinous process fractures [68, 116].

In underbody blast, therefore, spinal ligaments may play a role in limiting the angular displacement between vertebral bodies, and in affecting the pattern of bone failure seen. Axial loading of the spine in a neutral posture places the load acting anterior to the vertebral column, with the vertebral body in compression. The posterior ligaments are possibly placed under tension as the anterior vertebra is compressed to failure, perhaps leading to ligament failure or to failure of the ligament insertion with transverse process or spinous process fractures.

3.2.2 Disc properties

The disc, along with the facet joints, carries most of the compressive load the spine is subjected to during physiological activity with some load carried by the facet joints in certain postures [190]. The disc exhibits significantly rate dependent mechanical behaviour, which in physiological conditions means that it responds differently to short duration loads than to low level slow loading. Under compression, the disc is stiffer at high loads than low loads, thus providing flexibility at low loads and stability with high loads [190]. The disc is usually subject to compressive loads in physiological conditions, but when the vertebral column is in significant flexion the side of the disc on the concave aspect of the curve may be under tension.

During compressive loading, the nucleus pulposus behaves like a gelatinous mass, with rising pressure as load increases [5, 190]. As the pressure rises, the surrounding annulus builds circumferential tension along the collagen fibres [33] and the nucleus exerts load on the centre of the vertebral endplate [32, 159]. The tensile stresses in the annulus do not reach levels that could cause disc failure, even with very high compressive loads [190] and the endplate has been shown to fail before the disc under compressive load [33].

The behaviour of the disc at high strain rate loading is not well understood. Kemper *et al.* [103] loaded spinal motion segments from different levels in compression at different rates. It was shown that the compressive stiffness of the disc increased with loading rate, and that the effect was similar at different spinal levels. However, this study only tested three loading rates - 0.1 m/s, 0/2 m/s and 1 m/s - and did not control for posture or shear, so more research is needed.

3.3 Civilian Spinal Injury

Although this thesis does not focus on civilian spinal injuries, it is worth reviewing the most common injury patterns and mechanisms in the civilian literature.

A recent series from the United Arab Emirates [75] reports the epidemiology of spinal fractures. In this series, traffic collisions were the most common mechanism, followed by falls from height. Lumbar injuries were most common. However, this paper did not associate mechanism with pattern of injury.

Of more relevance to this thesis, Leucht *et al.* produced a review of civilian traumatic spine fractures that summarises the main injuries encountered in peacetime [110]. In this series, 562 patients with spinal fractures were reviewed. Falls from height over 2 m were the most common cause of injury, followed by road traffic collisions. There was a significant relationship between the mechanism of injury and the fracture pattern; falls lead to vertebral body compression fractures but road traffic collisions caused more flexion-distraction injuries. The paper notes that falls from “great height” were associated with flexion-distraction injuries but does not define the level of energy required. The paper also noted that falls from any height were associated with thoracolumbar junction injuries, but road traffic collisions led to cervical and thoracic injuries. Sports injuries were commonly at the cervicothoracic and thoracolumbar junctions, but were not associated with a particular injury pattern. This paper suggests that higher energy axial load injuries are perhaps associated with flexion-distraction rather than burst fractures, but there is insufficient detail to confirm this notion.

3.4 Summary

This section has described the anatomy of the spinal column in order to support understanding of descriptions of injury patterns in the spine. The current understanding of the fracture behaviour of the spine has been reviewed, including the difference in the way the spine fails at higher rates of loading such as may be seen in blast. At high strain rates, the behaviour of the component parts of the vertebral column is not well understood. The behaviour of the vertebral bone is better understood than that of the ligaments and discs, where detailed understanding is significantly lacking. It has been shown that the vertebral endplate fails before the intervertebral disc. The pattern of fracture can therefore be seen to relate to the mechanism of injury and loading rate. It may therefore be possible to derive the position of the spine at the point of fracture from the pattern of failure.

3.5 Spinal Fracture Classifications

There is a large number of classification systems for spinal injury. Most of these systems are published with the intent of simplifying clinical description and management of spinal fractures, but they all allow description of the anatomical disruption sustained. Given that each system describes injuries differently, they all bring different strengths and limitations and it may beneficial to use several systems in describing injuries for this thesis.

3.5.1 Cervical Spine

The unique anatomy of the cervical spine mandates the use of several classification systems to describe injuries at different levels. These describe injuries to the occipital condyles and atlantooccipital joint, the atlas and axis, and separately the subaxial cervical spine.

Levine and Edwards described C1 (atlas) fractures anatomically. Posterior arch fractures are associated with hyperextension injuries. Lateral mass fractures - injuries of the vertebra where the arches and pedicles join - are caused by axial load with lateral bending. Isolated anterior arch fractures are associated with hyperextension. Burst fractures secondary to significant axial load are referred to as Jefferson fractures [91] and describe three or more part fractures with disruption of the vertebral ring; a part is a distinct fragment of bone separated from its origin.

Axis fractures other than peg fractures are referred to as Hangman fractures. Levine and Edwards classified them based on a system by Effendi [58, 111]:

1. Minimally displaced pars interarticularis fracture (1a is identical with asymmetric fracture lines)
2. Displaced (>3 mm) pars fractures with flexion following hyperextension and axial load (2a fractures have flexion but no displacement)
3. Bilateral pars fracture with facet dislocation.

Subaxial cervical spine fractures were classified by Ferguson *et al.* [10]. This paper adopted a mechanistic description of cervical spine fractures, and the descriptions provide a spectrum of anatomical injury which is easily referred back to the mechanism:

- Compressive flexion (CF)
- Vertical compression (VC)
- Distractive flexion (DF)
- Compressive extension (CE)
- Distractive extension (DE)
- Lateral flexion (LF)

Although it is not possible to calculate reliably the magnitude of a force leading to fracture from the fracture pattern [10], it is feasible to extrapolate the direction of force and the mode of failure from the radiographic features. This is the strength of the Allen classification for this study. The key feature when considering the force vector is to identify the transition point between compressive failure and tension failure. This transitional axis, the point which marks the difference between modes of failure, is not necessarily identical to the point between compression and tension load but is the best analogue that can be identified from radiographic data [10].

The Ferguson-Allen system is subdivided into stages of severity. Compressive flexion fractures are divided into:

1. Blunting of anterior-superior vertebral margin with no failure of posterior ligaments;
2. Obliquity of anterior vertebral body and some loss of anterior or central height; possibly vertical fracture of body and inferior endplate changes;
3. Oblique fracture line passing from the anterior surface of the vertebral body, through the body, and in to the inferior subchondral plate, with a fracture of the “beak”;

4. Mild (<3 mm) displacement of the inferior-posterior vertebral margin in to the neural canal;
5. Displacement posteriorly in to canal, separation of facets, and increased interspinous distance indicating failure of the posterior ligamentous complex.

The major loading component in these fractures is an axial load, and multiple contiguous lesions suggest the cervical spine is in flexion at the time of injury [10]. The posterior failures in CF4 and CF5 injuries infer failure of the posterior elements in tension.

Vertical compression fractures are subdivided into:

1. Fracture of superior or inferior endplate with "cupping" deformity, with central failure of the endplate and no evidence of ligamentous disruption;
2. Fracture through both vertebral endplates with minimal displacement and possibly fracture lines through vertebral body;
3. Fragmentation of vertebral body with displacement in multiple directions around body, with or without ligamentous and vertebral arch disruption.

These injury patterns are associated with longitudinal compressive loading of the entire vertebral column and in these injuries the transitional axis lies posterior to the anterior elements of the vertebra [10].

Distractive flexion injuries combine a primary distraction force with secondary flexion:

1. Failure of posterior ligamentous structures with facet subluxation in flexion and increased interspinous distance;
2. Unilateral facet joint dislocation with possible small fracture of the articular surface;
3. Bilateral facet dislocation with 50% vertebral body displacement anteriorly with articular processes in contact or "perched";

4. Anterior displacement of full vertebral body width or grossly displaced vertebra.

Allen *et al.* [10] reported that the inferior vertebral body in motion segments affected by distractive flexion injuries often shows a compressive force applied to the anterior part of the vertebra, associated with distractive force posteriorly. The transitional axis lies in the middle third of the vertebral body, as seen in compressive flexion fractures, but there is not always a compressive injury to the lower vertebra.

Compressive extension fractures are associated with mechanisms which extend the head while compressing the cervical spine axially, such as a dive in to a shallow pool [10]. They are divided into:

1. Unilateral vertebral arch fracture, with or without vertebral body displacement. The arch fracture may be a linear fracture through the articular process, and ipsilateral pedicle and lamina fracture, or a combination of pedicle and articular process fracture.
2. Fractures of both laminae without evidence of other tissue failure.
3. Bilateral vertebral arch corner fracture of articular processes, pedicles, lamina, or a bilateral combination of these without vertebral body displacement.
4. Bilateral vertebral arch fractures with anterior displacement of the vertebral body.
5. Anterior displacement of the vertebral body by more than its width. In these fractures the posterior elements remain *in situ* while the anterior vertebral column translates anteriorly. There must be ligamentous failure at two levels, posteriorly between the supradjacent and fractured vertebra and anteriorly between the fractured and next inferior level. There is typically a shear fracture of the anterior-superior vertebral body below.

Vertebral arch fractures suggest a major injury vector directed towards the trunk, stressing the posterior elements in compression. Unilateral injury suggests that the stress is lateralised either by obliquity of the force vector or by a rotated posture of the cervical spine.

Distractive extension injuries are associated with falls from a height on to the face, leading to spine extension [10]. They are categorised into:

1. Failure of the anterior ligamentous complex or a transverse undisplaced body fracture. There may be a fracture of the adjacent vertebral body margin. The disc space may be widened.
2. Failure of the posterior ligamentous complex with displacement of the upper vertebral body posteriorly in to the canal.

Lateral flexion injuries are caused by a laterally directed force leading to the head flexing towards the shoulder. They are often associated with brachial plexus injury [158]. Allen *et al.* [10] subdivided them in to two:

1. Asymmetric compression fracture of the body with ipsilateral vertebral arch fracture. Possible compression failure of the articular process or comminution of the corner of the vertebral arch. There may be a vertical fracture of the body.
2. Lateral asymmetric compression of the vertebral body and either ipsilateral vertebral arch failure or contralateral ligamentous failure under tension with separation of the articular processes.

It is interesting to note that in this classification system, where multiple injuries occur concurrently, DF lesions are one level above CF lesions, which are again one level above VC injuries. The transitional axis also lies progressively more dorsal at lower levels [10]. Multiple fractures with similar patterns were only seen in Allen's series with compressive failure. However, these descriptions are based on civilian injury mechanisms. This system is quite complex and is not often used in clinical settings, but is common in research papers because of its flexibility and mechanistic relevance, so is suited to military injury.

3.5.2 Thoracic and Lumbar Spine Injury Classification

As in the cervical spine, thoracolumbar fractures have been described with a variety of classification systems, each with a slightly different way of describing similar injuries.

McAfee *et al.* [121] described a simple anatomical and mechanistic system which has been used in several review papers of military spinal injuries [149] [57]. Fractures were divided into axial compression, axial distraction, and translation, all based around the forces injuring the middle column. This gave several descriptive fracture patterns. Wedge compression fractures cause isolated failure of the anterior column. A stable burst fracture shows compressive failure of the anterior and middle columns, but no posterior column injury. Unstable burst fractures involve posterior column disruption. A Chance fracture [38] is a horizontal avulsion injury of the vertebral body about an axis anterior to the anterior longitudinal ligament – it is a pure distraction injury with no anterior column fracture. Flexion distraction fractures are characterised by a motion axis posterior to the anterior longitudinal ligament and demonstrate compressive failure of the anterior column and distraction failure of the posterior column. Translational injuries are characterised by disruption of the alignment of the neural canal, implying failure of all three columns in shear. The McAfee system is universally understood and used in most of the papers that describe blast injury in the spine.

Magerl *el al.* [118] described another classification system for thoracolumbar injuries (Figure 26). This mechanistic description drew on the work by McAfee *et al.* [121] and Ferguson *et al.* [62] to relate the mechanism of injury to the fracture pattern produced, classifying fractures broadly in to vertebral body compression (Type A), anterior and posterior element injury with distraction (Type B), and rotational injuries (Type C). The major fracture types are further subdivided and allow a detailed description of each injury. The system is detailed in Table 2, Table 3, and Table 4. This is a complex system which shares some terminology with other classifications but may allow a more detailed review of the injury patterns seen in blast. None of these classification systems reports transverse or spinous process fractures. Although these injuries are of themselves relatively insignificant, they may give useful information in inferring the mechanism of a given blast injury. The next chapter describes the patterns of injury in published military spinal trauma experience.

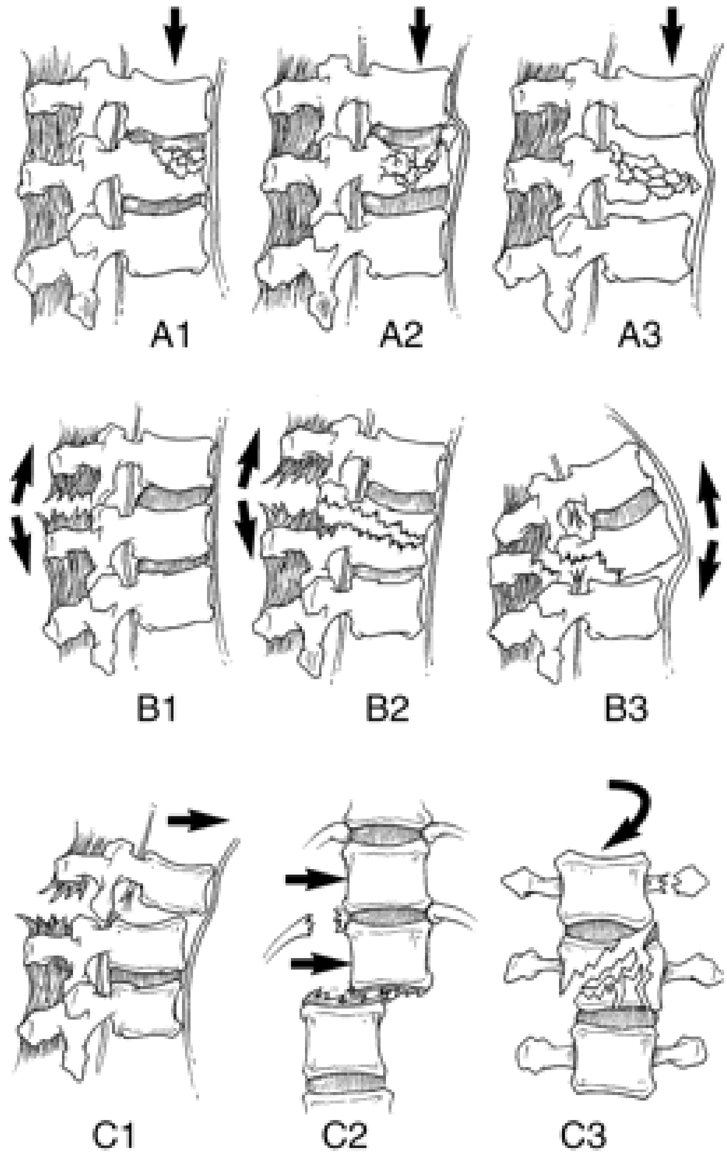

Figure 26: Magerl classification of thoracolumbar fractures. Reproduced with permission from Mirza with permission from Walters Kluwer. Promotional and commercial use of the material in print, digital or mobile device format is prohibited without the permission from the publisher Wolters Kluwer Health. Please contact lwwjournalpermissions@wolterskluwer.com for further information [125]

A1	**Impaction fractures**			
	A1.1	Endplate impaction		
	A1.2	Wedge impaction fractures	1	Superior
			2	Lateral
			3	Inferior
	A1.3	Vertebral body collapse		
A2	**Split fractures**			
	A2.1	Sagittal split fracture		
	A2.2	Coronal split fracture		
	A2.3	Pincer fracture		
A3	**Burst fractures**			
	A3.1	Incomplete burst fracture		
			1	Superior
			2	Lateral
			3	Inferior
	A3.2	Burst-split fracture		
			1	Superior
			2	Lateral
			3	Inferior
	A3.3	Complete burst fracture		
			1	Pincer burst
			2	Flexion burst
			3	Axial burst

Table 2: Magerl Type A fractures [118].

B1	**Posterior disruption, predominantly ligamentous: flexion-distraction injury**		
	B1.1	With transverse disruption of the disc	
		1	Flexion-subluxation
		2	Anterior dislocation
		3	1 or 2 with articular process fracture
	B1.2	With Type A fracture of body	
		1	Flexion-subluxation and fracture
		2	Anterior dislocation and fracture
		3	1 or 2 with articular process fracture
B2	**Posterior disruption, predominantly osseous flexion-distraction injury**		
	B2.1	Transverse bicolumn fracture	
	B2.2	With transverse disruption of the disc	
		1	Disruption through pedicle and disc
		2	Disruption through pars interarticularis and disc
	B2.3	With Type A fracture of vertebral body	
		1	Fracture through pedicle and body
		2	Fracture though pars and body
B3	**Anterior disruption through disc (hyperextension shear injury)**		
	B3.1	Hyperextension-subluxation	
		1	Without posterior column injury
		2	With posterior column injury
	B3.2	Hyperextension-spondylolysis	
	B3.3	Posterior dislocation	

Table 3: Magerl Type B injuries [118].

C1	**Type A injuries with rotation**		
	C1.1	Rotational wedge fracture	
	C1.2	Rotational split fractures	
		1	Sagittal
		2	Coronal
		3	Pincer
		4	Vertebral body fracture
	C1.3	Rotational burst fractures	
		1	Incomplete
		2	Rotational burst-split
		3	Complete rotational burst
C2	**Type B fractures with rotation**		
	C2.1	B1 injuries with rotation	
		1	Rotational flexion subluxation
		2	Unilateral articular process fracture
		3	Unilateral dislocation
		4	Rotational anterior dislocation
		5	Rotational flexion/subluxation
		6	Unilateral dislocation and Type A
		7	Rotational anterior dislocation and Type A
	C2.2	B2 injuries with rotation	
		1	Rotational transverse bicolumn
		2	Unilateral flexion spondylolysis
		3	2 with Type A fracture
C3	**Rotational shear injuries**		
	C3.1	Slice fracture	
	C3.2	Oblique fracture	

Table 4: Magerl Type C injuries [118].

4. Military Spinal Injuries in the Published Literature

This thesis has reviewed the effect of blast on a vehicle and the biomechanics of the spine in trauma. It has also discussed the importance of injury prediction models in designing vehicles to mitigate the effect of blast, and shown that these models require an understanding of the behaviour of the spine during a blast event. This chapter reviews the published literature describing spinal injury in warfare.

4.1 Historical Published Data

Historically, publications detailing injury patterns in warfare have focussed on thoracoabdominal injuries, cause of death, and limb trauma. Although spinal injuries in warfare were reported as long ago as the Egyptian era [69, 163], publications prior to the Gulf conflict of 2003 pay little regard to spinal injuries. As an example, in 1980, Owen-Smith described the injury patterns in 2,000 soldiers but did not specifically mention spinal fractures, though patients with paraplegia and paraparesis secondary to spinal cord injury were reported [137]. This paper did not record the level of injury. Interest in spinal injury in military patients developed with progression of the campaign in Afghanistan. Eskridge *et al.* described 4,263 injuries in soldiers in the Iraq conflict in some detail but described the incidence of spinal injury simply as “low” [59].

This chapter reviews the existing literature to identify the demographics of the military trauma injury population, in particular to identify the epidemiology of spinal fractures in warfare, and to examine the mechanistic classification and distribution of such injuries. The reviewed papers are discussed below and their findings summarised and tabulated. Papers describing the patterns of injury encountered in warfare do not always detail the mechanism of injury or whether the victim was mounted (in a vehicle) or dismounted. Papers that specifically deal with mounted blast victims will be considered separately as this thesis will focus on the mechanism of spinal injury in mounted victims.

4.2 Literature Review Methodology

A Pubmed, Google Scholar, and Embase search was carried out to identify papers relating to military spinal injury, using the terms “spinal fracture blast”, “blast injury spine”, and “ballistic injury spine”. MeSH terms found and used are shown in Annexe 1. Additional records were identified by reviewing the references in each paper. Papers that were not available in English were excluded. Initial exclusions were of papers with no mention of spinal fractures (4) and those which appeared to use the same patient population (3). The search was initially carried out in October 2013 and repeated in May 2015.

Papers were reviewed to identify details of spinal injuries in warfare including blast and ballistic mechanisms. Data was sought for the injuries associated with each of these mechanisms and which vertebrae were injured. The summary measure for this review was the number of fractures at each spinal level reported in each paper. No meta-analysis was possible due to the limitations of the majority of publications, but summary measures are reported here. The review methodology is summarised in Figure 27. Select papers describing spinal injuries in the whole military population will be reviewed, focussing on those which identify both mechanism and injury.

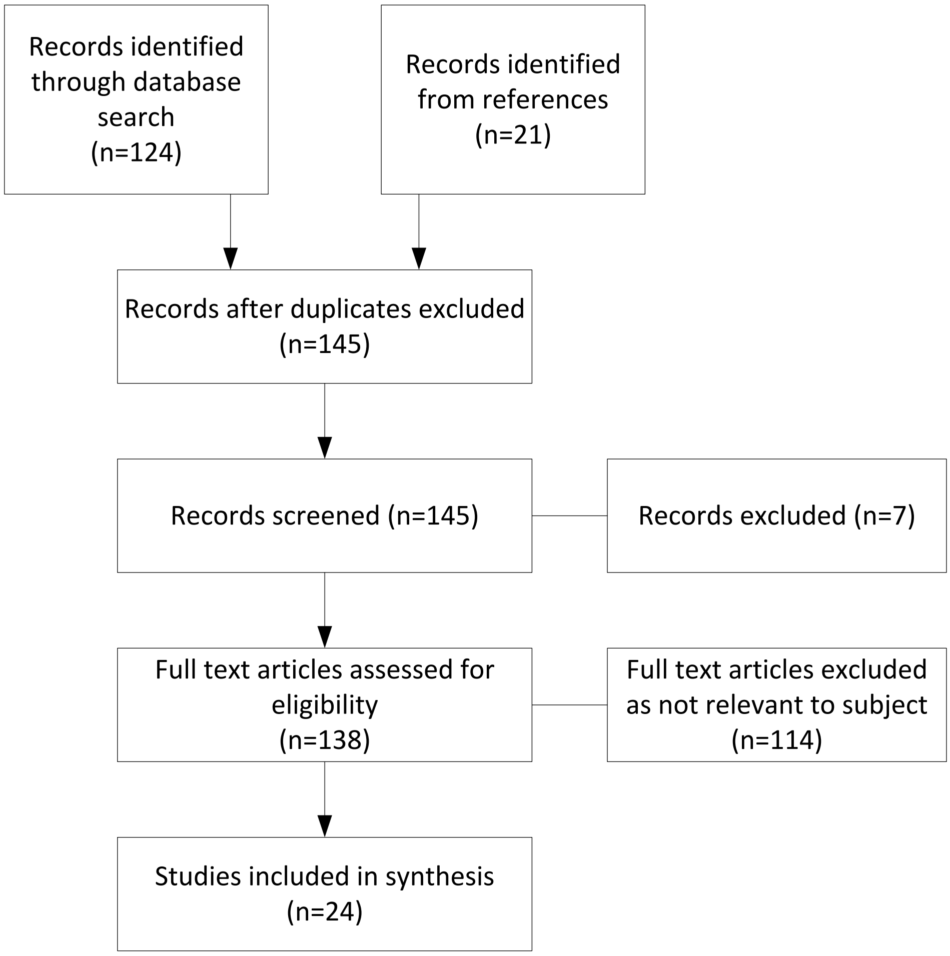

Figure 27: Literature review summary according to PRISMA guidelines [126].

4.3 Results

4.3.1 Demographics and Cause of Spinal Injury in War

It is clear from the recent literature that blast injury causes the bulk of spinal injuries in current conflicts and that the patients affected are young; Table 5 lists the nine papers which report specific mechanisms of injury and patient demographics.

	Mean Age	Male %	Female %	All Blast %	Gunshot %	Motor vehicle collision%	Air crash %	Fall from Height %	Other and Unknown %
Bell [16]	27	98	2	56	14	7	3		7
Belmont [18]	25.8	98.5	1.5	75	20	3			3
Blair [27]	26.4	98	2	56		27		6	11
Blair [26]	26.5	98	2	56	15	25	5		
Comstock [44]	29.3	99.5	0.5	72	7	9		5	7
Possley [145]	26.5	98.4	1.6	53					
Schoenfeld [164]	27.8	92	8	83	3	3	9		
Schoenfield [170]	26.6	98	2	67	15		11		7
Schoenfield [169]	26.6	99	1	75	14.8	7.8			
Unweighted Mean	27	98	2	66	13	12	7	6	7

Table 5: Demographics and Cause of Injury in published data for military spinal fractures. Unweighted because not all papers provided enough data for weighting.

4.3.2 Distribution of Military Spinal Injuries

The United Kingdom and the United States have both established registries of trauma patients in conflict over the past decade. These registries have been used to provide data for several UK and US papers examining the distribution of injuries in Iraq and Afghanistan. There have also been some focussed case series.

Eardley *et al*. [57] reviewed military spinal injuries from both blast and ballistic injury mechanisms. Injuries were classified according to mechanism (explosion, gunshot, and non-ballistic) and fracture pattern. Of those patients reviewed whose injury was a result of blast alone, the majority were shown to have unstable lumbar spine burst fractures. Unstable lumbar burst fractures suggest a higher energy insult than stable injuries. Unfortunately, comparable papers do not all separate stable and unstable injuries.

Belmont *et al*. reviewed the overall wounding patterns in 6,092 casualties in US forces between 2005 and 2009 [18]. 17,177 musculoskeletal wounds were identified with 82% caused by blast. 797 spinal fractures were reported, of which 595 were due to blast. The type and level of injury were not mentioned. Schoenfield *et al*. [169] described the same 797 patients with spinal injuries in a different paper, focussing in particular on their spinal trauma. 74.5% of those patients had spinal injuries caused by blast. The overall incidence of spinal injuries in soldiers deployed during this time period was 4.4 per 10,000 soldiers, with 11.1% of combat casualties suffering a spinal injury. Multi-level spinal fractures occurred in 31% of blast patients. However, the paper does not describe the individual vertebral levels injured, nor does it separate blast from other causes of injury at each level. It is also not clear whether the patients with injuries ascribed to blast were exposed to dismounted blast and fragmentation injury or were mounted in vehicles.

A review of casualties in a single US Army Brigade Combat Team [164], also from Schoenfeld's group but apparently using a different data set, focused on spinal injuries. Seventeen fractures were identified in eight soldiers with explosions as the most common mechanism. The paper described 31 spinal injuries including low energy soft tissue injuries which are not considered in this review. Cervical spine fractures were the most common with two fatal cervical injuries. Individual vertebral levels were identified for the blast injury patients and have been included in the analysis later in this section.

Blair *et al*. [26] described 598 spinal column injuries in 10,979 combat casualties between 2001 and 2009. 56% of those injuries were caused by explosive mechanisms and although the paper describes the fracture patterns in detail, it does not report which mechanism lead to which fracture pattern. Most blunt injuries were lumbar and the most common classification, other than transverse process, was compression. However, this paper groups all blunt injuries together, including non-blast casualties. The same group of patients is reported in a second paper by the same lead author [27] to compare blunt and penetrating trauma, noting that spinal cord injury is more common in penetrating injuries.

Lehman *et al.* [108] described a series of patients with lumbar burst fractures sustained in Iraq and Afghanistan, reviewing thirty-two patients with 39 fractures. No patient in this study had an isolated single vertebral burst fracture. 34% of patients were injured as a result of IED strike. The paper does not report the cause of individual injuries, nor does it mention whether the IED patients were mounted or dismounted. However, it is worth noting that in civilian motor vehicle, sports, and osteoporotic vertebral fractures, most burst fractures occur above L2 [20, 86, 106]. This paper suggests that soldiers wearing body armour have sufficient additional support at and around the thoracolumbar junction that the transition zone between the rigid thoracic and flexible lumbar spine, normally responsible for the high incidence of thoracolumbar injuries in civilian trauma [110], is moved caudally.

The Polish detachment to Afghanistan published a case series of their injuries in the campaign [202]. This reported the majority of spinal injuries to be compression fractures at T12-L1, which was attributed to the seated posture in a vehicle, but further details were not reported.

It can be seen from the summary of these papers in Table 6 that the great majority of military spinal fractures are in the lumbar region. Figure 28 shows the number of fractures in the reviewed literature at each vertebral level in all war spinal injuries; it is clear that the majority of fractures occur in the junctional regions of the spine. However, as has been shown, the published data do not provide much detail in their descriptions of military spinal injuries.

Study	Cervical		Thoracic		Lumbar	
Bevevino [23]. Spinal injuries in combat amputees.	5	(6%)	15	(18%)	62	(76%)
Bilgic [24] Case report of lumbar burst fracture due to anti-personnel mine.					1	(100%)
Blair [26] US Casualties 2000-2009.	319	(18%)	591	(33%)	857	(49%)
Comstock [44] Canadian casualties.	6	(13%)	15	(33%)	25	(54%)
Davis [51] Injuries on the USS Cole.	2	(17%)	8	(73%)	1	(8%)
Eardley [57] Review of British military spinal trauma	2	(5%)	14	(32%)	28	(64%)
Lehman [108] Review of the "Low lumbar burst fracture"					39	(100%)
Possley [145] Review of spinal injuries in IED strike.	279	(17%)	543	(34%)	787	(49%)
Schoenfeld [170] Review of fatal injuries in US troops.	704	35%	731	36%	579	29%
Schoenfeld[164] Review of injuries in a single US unit.	4	(40%)	2	(20%)	4	(40%)
Schoenfeld[169] Review of US casualties 2005-9.	231	(22%)	300	(28%)	522	(50%)
Turegano-Fuentes [185] Review of injuries in the Madrid train bombings.	6	(29%)	15	(71%)		
TOTALS	1558	(23%)	2234	(33%)	2905	(43%)

Table 6: Overview of military spinal fracture distribution from all causes in all reviewed papers showing the clear trend towards lumbar spinal injuries.

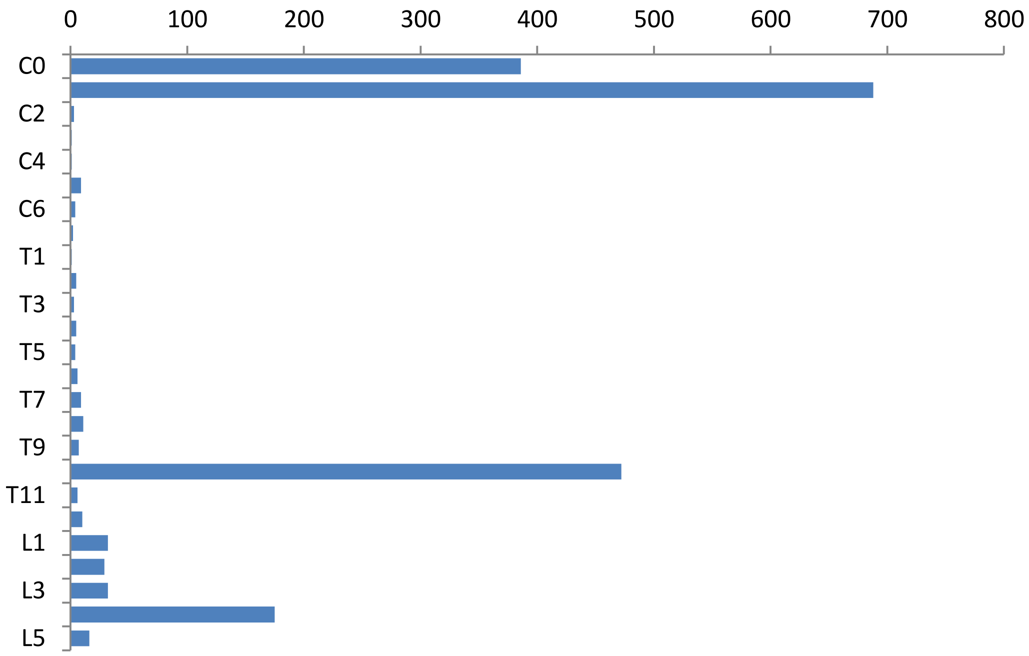

Figure 28: Injury distribution by vertebra showing the number of fractures at each level. All mechanisms in all reviewed papers for both survivors and fatalities in mounted and dismounted blast.

4.3.3 Blast Injury Distribution in the Published Literature

Several papers report the patterns of injury seen in military explosive casualties but do not describe the spinal injuries in detail. For example, Schoenfeld *et al.* [166] reported 4048 cervical fractures in the US military between 2000 and 2009, with a detailed description of the demographics of the study population but no mention of the level of the fractures or the incidence of blast injury in those patients. The majority of fractures in blast are lumbar or thoracic (Table 7) [23, 24, 26, 44, 51, 57, 108, 145, 164, 169, 170, 185]. However, many of these papers group mounted and dismounted victims together, with their disparate mechanisms of injury.

	Cervical		Thoracic		Lumbar	
Bevevino [23] Spinal injuries in combat amputees	5	(6%)	15	(18%)	62	(76%)
Bilgic [24] Case report of anti-personnel mine injury			1	(100%)		
Eardley [57] Review of British military spinal trauma	2	(5%)	14	(32%)	28	(64%)
Helgeson [81] Lumbosacral dissociation in blast injury					23	(100%)
Ragel [149] Thoracolumbar injuries from anti-vehicle blast			4	(24%)	13	(76%)
Schoenfeld [164] Review of injuries in a single US unit	3	(27%)	1	(14%)	3	(43%)
Turegano-Fuentes [185] Review of injuries in the Madrid train bombings	6	(29%)	15	(71%)		
Total	16	(9%)	49	(28%)	107	(62%)

Table 7: Spinal injury distribution by level from blast mechanisms – survivors and fatalities, both mounted and dismounted, but gunshot and other mechanisms are excluded.

4.3.4 Mounted and Dismounted Blast

Victims of blast on foot (dismounted) are likely to be subject to a different, and more varied, mechanism of injury than in-vehicle (mounted) victims.

The Balkans conflict also produced significant numbers of victims of dismounted blast [90]. In 1998 the neurosurgical department in Split reported 96 patients with spinal fractures and cord injury from blast, gunshot, and fragmentation mechanisms. 12 isolated fractures, 68 fractures with cord injuries, and 6 fracture-dislocations were reported. The level of each fracture was not made clear, but the majority of injuries were lumbosacral. In a report of the 2004 Madrid train bombings, 512 victims had 21 spinal fractures [185]. Most of these were in the upper thoracic region. Caution is required when comparing these injuries to mounted or recent military dismounted injury patterns as the victims were not wearing body armour, helmets or other heavy military equipment. Additionally, the victims would have been exposed to blast from different directions and in different seated postures, therefore applying a variety of confounding mechanisms to any useful comparison.

The Canadian experience of spinal trauma was also reported in 2011 using a national trauma registry [44]. Of 29 patients with a spinal fracture, there were 46 fractures due to blast in 23 patients with the majority in the lumbar region. It is not possible to separate mounted and dismounted patients from this paper.

Possley, in 2012, compared mounted and dismounted blast injuries from US data [145]. In this series, victims of blast with and without vehicle protection were compared, and mounted blast victims before and after the introduction of heavier up-armoured vehicles were also compared. It was shown that the incidence of fractures in dismounted victims was similar to that in mounted victims with 1,347 fractures in 404 dismounted victims and 472 fractures in 145 mounted victims. Surprisingly, later vehicle designs aimed at reducing the injury burden were associated with more severe fractures than earlier vehicles with 27% of spinal fractures in the early group being classified as "major" compared to 34% in the later group. This may be confounded by an increased survival rate in later vehicles or the use of larger devices in attacking those vehicles, but this is not reported. Gunshot wounds were more common in the dismounted group, but this is to be expected given the protection from gunshot afforded by an armoured vehicle.

4.3.5 Solid Blast

Solid blast is the injury mechanism in which an explosive attack on a vehicle transmits significant indirect force to the occupants through the vehicle itself. It is solid blast which is analogous to blast injury in mounted victims; this thesis will focus on mounted blast as these are likely to be the injuries most amenable to understanding and mitigation.

Ragel *et al.* [149] focussed on survivors of IED strikes on armoured vehicles, describing a solid blast injury pattern in 12 patients with 17 thoracolumbar fractures with 38% of the fractures described as flexion-distraction injuries. In these injuries there is either a horizontal fracture line through the pedicle or a posterior soft tissue injury associated with minimal anterior height loss and possibly increased height of the posterior vertebral body. This paper, unlike McAfee's original description [121], did not separate flexion-distraction injuries and Chance injuries, which are pure distraction injuries with no compression component [38]. Even in this series, however, the majority of injuries were burst pattern fractures with most occurring at L1, and not all papers support the notion that there are more flexion-distraction injuries in mounted blast victims than in other groups. Flexion-distraction injuries and Chance fractures are grouped together in the summary below as some papers do not divide them.

In the military population, Ragel suggested several different mechanisms for these injuries when soldiers are exposed to underbody blast [149]. These include flexion of the spine as a result of the legs being forced upwards by floorpan deformation with the torso held in place by a seat harness, as shown in Figure 29. In order for the spine to rotate with leg movement, the pelvis must presumably also rotate into flexion; the seat harness provides a pivot point. This would be compatible with the mechanisms of injury noted in solid blast calcaneal fractures [154]. An alternative mechanism suggested is flexion of the spine as a result of body armour and other equipment [108].

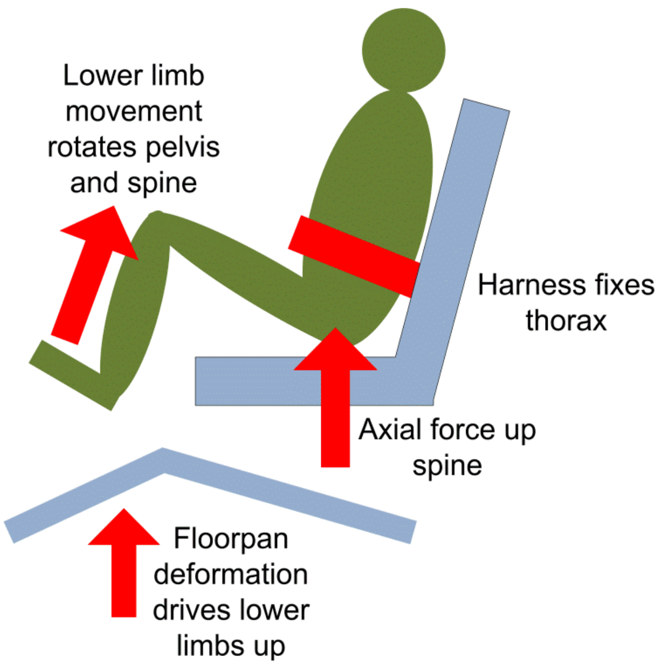

Figure 29: Effect of underbody blast on seated victim suggested by Ragel. The deforming flooorpan throws the lower limbs upwards, rotating the pelvis and lumbar spine about the fixed torso. [149].

Another injury pattern that is postulated to be more common in blast trauma is lumbosacral dissociation [81]. In the civilian population, fractures of the sacrum with associated high-grade traumatic spondylolisthesis are rare and usually due to a motor vehicle collision or a fall from significant height [21, 193]. It has been suggested that these injuries in the civilian population are related to significant axial loading and this mechanism would be compatible with the high rate axial load injuries seen in underbody blast.

Extracting the fracture patterns from these papers, where possible, demonstrates that compression and burst fractures are the most common mechanistic descriptions of fractures in blast injury (Table 8). Transverse process fractures are common but are disregarded in some papers as they are not of themselves significant. They are, however, considered to be a marker of high energy injury [140]. The figure and table presented here combine descriptions of “stable burst” and “unstable burst” from Eardley *et al*. [57] as not all papers subdivided these fractures. Thoracic and lumbar burst and compression fractures, although not identical, are also grouped together as some papers do not divide them. The aggregate figures in Table 8 therefore may not be wholly representative, given the different priorities of each paper, but it is felt that this is the best available data to compare.

Although the published data provide limited detail with respect to both mounted and dismounted blast injury in the spine, some patterns do begin to emerge. In both mounted and dismounted groups, the majority of injuries are thoracolumbar. Some particular patterns are suggested to be more common in blast, such as low lumbar burst fractures, flexion-distraction injuries, and lumbosacral dissociation. These suggest that the predominant cause of blast related spinal fractures is probably axial load. However, a comprehensive review of blast related spinal injuries is needed if we wish to clarify these possibilities.

	Odontoid Peg	Facet Fracture Dislocation	Lamina	Transverse process	Compression	Burst	Flexion-Distraction and Chance	All Sacral
Bevevino[23]		1		45	18	8		
Comstock[44]			1	14	7			
Eardley[57]					6	19	4	
Helgeson [81]								24
Ragel[149]					7	3	5	
Schoenfeld[164]	1			5	3	1		
Total	1	1	1	64	41	31	9	24
%	1%	1%	1%	37%	24%	18%	5%	14%

Table 8: Fracture patterns in blast patients in the whole published literature. Most fractures are due to axial loading, or combine anterior compression with posterior axial loading.

4.3.6 Patterns in Fatal Blast Injury

Several papers do not separate fatal from non-fatal injuries, or do so with insufficient detail to identify any mechanisms or injuries which might be associated with fatality. Understanding whether a particular pattern of injury is associated with fatality might support understanding mitigation strategies to prevent these injuries, therefore saving lives. The most detailed papers are summarised at Table 9 and the overall distribution is shown in Figure 30. Even so, these papers do not all separate blast injury from other mechanisms.

One paper specifically reviewed spinal injury patterns in fatal incidents [170]. This paper described 2,089 cases of spinal trauma from a series of 5,424 patients killed in Iraq and Afghanistan between 2003 and 2011. There was a high incidence of cervical spine fractures and fracture dislocations, with 378 spinal dislocations including 223 spinal cord transections. 52% of these patients had at least one cervical injury, with 686 C1 injuries. There were also 285 subaxial cervical injuries, but the vertebral levels involved were not reported. This is in marked contrast to the bulk of published literature which shows a higher incidence of more caudal injury. The paper reports that cord injuries were associated with gunshot wounds and atlantooccipital trauma was associated with blast, but does not report the numbers of blast patients in each group. There was a significant incidence of concomitant pelvic, head, brain, thoracic, and abdominal trauma in these patients, but the cause of death was not itself recorded.

The suggestion that there is a correlation between intracranial injury and spinal injury may be borne out by a series of patients seen in a neurosurgery service following injury in Iraq [16], in which 40 out of 428 patients seen by neurosurgeons had spinal column or cord injury. However, the paper does not report in detail the nature or level of those injuries.

One paper specifically reviews the distribution of injury in fatal solid blast from a single attack. The American ship USS Cole [51] was attacked with explosives loaded into a small boat while at port in October 2000, sustaining significant structural damage and exposing casualties to a solid blast mechanism. The paper does not detail the exact location of casualties compared to the source of blast. There were 17 fatalities with 11 fractures. One survivor was reported to have had spinal column fractures but the level of fracture was not included. All these patients had multiple severe injuries other than the spinal and were judged to have been unsurvivable.

	Cervical		Thoracic		Lumbar	
Davis [51]	2	(17%)	8	(73%)	1	(8%)
Schoenfeld [164]	2	(100%)				
Schoenfeld [170]	1095	(41%)	924	(35%)	638	(24%)
Total	1099	(41%)	932	(35%)	639	(24%)

Table 9: Spinal injury distribution in fatal casualties for all military mechanisms. Cervical spine injuries are more common in fatalities than in the whole blast victim population.

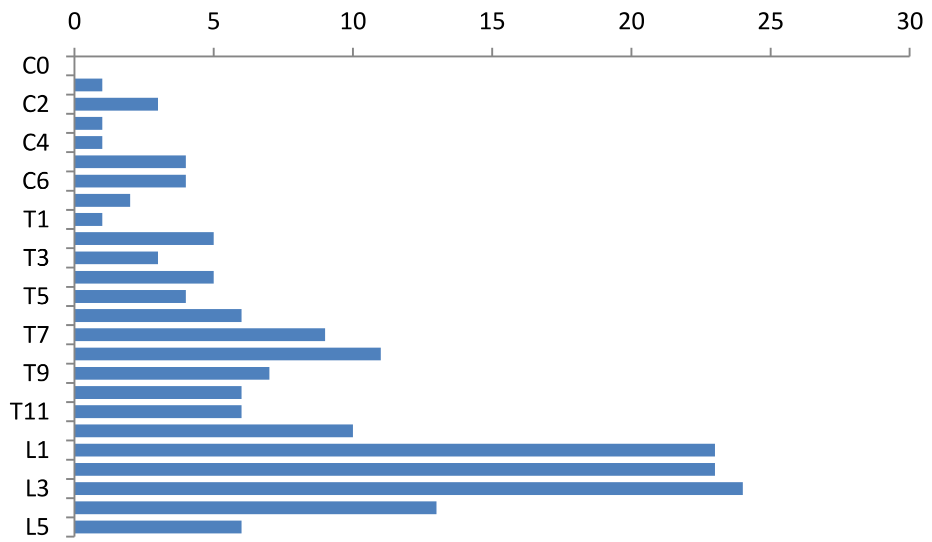

Figure 30: Injury distribution by vertebra in blast victims in all papers, both survivors and fatalities. Most fractures are lumbar. There is a peak in the mid-thoracic spine at the apex of the thoracic curve.

4.3.7 Neurological Injury

Spinal cord and nerve injuries are not the focus of this thesis but are likely to have a significant impact on the clinical effect and outcome of military spinal injury. They will therefore be discussed briefly.

Hunter *et al*. [87], in a 1941 discussion on the problem of blast injury, first noted haemorrhage surrounding nerve roots in the spinal canal of animal models exposed to primary blast. It is therefore possible that compromising neurological injury might be present in the absence of spinal column fracture or dislocation.

Several papers reporting the incidence and patterns of spinal injury in warfare do not separate blast patients from those caused by other mechanisms such as gunshot, motor vehicle collisions, and falls from height. This is critical, as Possley *et al*. [145] found that spinal cord injury was more common in dismounted than mounted troops with the former more likely to have been injured by gunshot, and Blair *et al.* [27] noted a higher incidence of cord injury in penetrating than blunt mechanisms. Blair's paper, however, included mechanisms other than blast.

Schoenfeld *et al*. [169] reported spinal cord injuries in 9%, and nerve root injuries in 3%, of patients with spinal injuries from all military mechanisms during 2005-2009. In the same group's series of fatalities with spinal injury [167], there was a significant incidence of spine transection (223 patients, 9.6%) and cord injury (834 soldiers, 40%). The most common level of cord injury was C1 (113 patients, 13.5%).

4.3.8 Associated Limb, Head and Visceral Injuries

Although this thesis focuses on spinal injury mechanisms, it is useful to appreciate the significance of associated organ and limb injuries as these may help describe the axis of an incurring force or suggest an associated injury mechanism, such as the head striking an object causing a skull fracture.

Eardley *et al.* [57] noted concurrent non-spinal injuries in 60% of spinal injury patients. This paper did not find a significant correlation between spinal fracture level and particular organ injuries. Blast injuries were found to be associated with a significant probability of associated extremity injury.

Bevevino *et al*. [23] reported the incidence of spinal fractures in combat related amputees. Twenty-nine of the 226 amputation patients in this series also had spinal fractures, with 82 fractures in total. Most (62, 76%) were in the lumbar spine and were transverse process or compression fractures. The paper does not specifically record which patients were involved in mounted or dismounted blast incidents, though it implies that all the injuries were due to blast. A further review [141] focuses on patients with significant associated injuries and lists 1,690 fractures, most of which are lumbar; it notes that 76% of patients sustained multiple fractures and 62% of injuries were a result of blast. This paper appears to draw on the same series of patients as another paper [26] by the same group and has therefore not been used in the overall analysis.

In victims of fatal blast injury [170], it is not surprising to note that there is a very high incidence of associated injuries. In this series, patients with spinal injuries killed by all mechanisms had a mean of 20.6 associated injuries. 70% had associated head and neck injuries and abdominal and extremity injuries were present in more than half.

4.3.9 Outcomes and Surgical Management in Wartime Spinal Injury

The Edwin Smith Papyrus [69] records the futility in early medical history of treating spinal injuries associated with paralysis, advising physicians not to attempt to treat such injuries. Little was then published with regard to the outcomes of spinal injury until the Second World War, which provides some data. The shortest follow-up was published by Cozen [47] and reported improvement in symptoms within a few days for patients exposed to blast and complaining of back pain, but did not record an anatomical diagnosis.

One long term series was published following Second World War injuries [162]. This series reviewed 56 American soldiers with penetrating spinal wounds from explosive fragment or gunshot. Treatment was by laminectomy when there was progressive neurological abnormality or evidence of metallic or bony fragments in the spinal canal on plain radiographs, with follow-up to a maximum of 40 months. At final review, 4 of the 19 patients who were paraplegic at the time of injury had some recovery and 22% of all patients with neurological deficit made a complete recovery. 36% of patients with a neurological deficit immediately following injury made no recovery. However, this did not include closed fractures so is not comparable to the solid blast injuries on which this thesis is focussed.

Neurological outcomes in patients with penetrating ballistic spinal trauma from Afghanistan were reported by Blair *et al*. [27]. 17 out of 23 (74%) blunt trauma patients with a neurological deficit showed improvement at follow-up compared with 19 of 34 (56%) patients with penetrating wounds.

Specific case reports of individual patient outcomes are rare. Kang *et al.* [94-97] reported several specific cases. One patient is described with an L5 burst fracture following exposure to blast from an IED associated with bilateral transfemoral amputations and normal neurological function in the residual limbs, but 50% occlusion of the spinal canal. He was treated with L4 to S1 fusion and achieved a pain-free outcome despite needing steroid injection for radicular pain. In this case, surgery was advocated despite the lack of neurological compromise in order to facilitate rehabilitation.

Neurological injuries associated with lumbosacral dissociation or other spinal dislocations might reasonably be expected to be associated with a worse outcome. One case of an open lumbosacral dissociation [95] associated with bilateral transfemoral amputations leading to long term bowel and bladder disturbance. A better outcome was reported by the same authors [96] in a closed lumbosacral dissociation injury, also associated with bilateral lower limb amputations. In this case the fracture was stabilised with L4 to ileum posterior fusion and the patient was able to mobilise with prostheses and no reported neurological deficit.

Complications following treatment of military spinal injuries have been reported in one US series [146]. The overall complication rate following spinal trauma was 9% with a high rate of multiple complications. Wound infections, venous thrombosis and cerebrospinal fluid leak were the most common complications and patients injured in dismounted mechanisms were at higher risk.

As some of the review papers have described gunshot wounds, it is worth describing them briefly. Gunshot wounds to the spine have been reported in civilian and military populations. Kang *et al.* [97] reported one case of a large calibre round in the thoracic spine. The patient had no residual lower limb motor function following laminectomy and dural repair. A civilian series of spinal gunshot wounds [30] suggests that most such injuries in the civilian population are stable and require surgery only for retained fragments or evolving neurological deficit. Another series of gunshot wounds in the cervical spine [122] suggests that unstable fractures are unlikely in the absence of neurological deficit.

4.4 Summary

The incidence of spinal injury in warfare may have increased in recent conflicts. While spinal injuries remain less common than limb injuries, they have the potential to cause significant disability and therefore are worthy of study. There are few publications relating to spinal war injury, however, until the recent interventions in Iraq and Afghanistan. The papers considered here are listed in Annexe 2.

The available data suggest that most spinal injuries in victims of blast are in the thoracolumbar and lower lumbar spine. There appears to be a significant incidence of unstable burst fractures. There may be a different pattern of wounding in dismounted and mounted incidents, but this is not clear. These patterns both appear to be different to that seen in the civilian population with more unstable and low lumbar burst fractures. Several possible mechanisms for this have been proposed but none has been proven. It seems likely that as the seat and legs are accelerated upwards, the pelvis and lumbar spine flex, then the spinal column compresses sequentially with some degree of flexion as it does so, leading to flexion-compression injuries.

Solid blast has received attention in recent literature with regard to lower limb injury. However, case series suggest that there is a significant rate of spinal injury in these patients and the mechanisms and patterns of wounding have not been explored in detail. Published data often do not separate spinal fractures from other spinal column injury and do not always report neurological injury or outcomes. They often do not separate the mechanisms of injury in sufficient detail to enable analysis in depth.

Cervical spine fractures appear to be more common in blast fatality. However, many papers do not separate fatalities from survivors, and those that focus on fatalities do not necessarily separate blast from other mechanisms of wounding. It is unclear whether this trend is significant, given the paucity of detail in the published literature. If it is indeed true, then identifying the specific injury patterns might allow changes in vehicle and equipment design to reduce the incidence of such fatal injuries.

This chapter has reviewed the literature with respect to spinal injuries in warfare, with a focus on mounted blast victims. It has shown that the majority of injuries involve the thoracolumbar spine, but that the published papers lack the detail necessary to derive mechanistic information. Injury patterns may be used to derive a mechanism of injury. The next chapter identifies the distribution of injury in UK blast victims and evolves a mechanistic explanation of those injuries.

5. Injury patterns in UK blast victims and their mechanistic implications in underbody blast[1]

Previously, this thesis has identified the patterns of spinal injury described in the - mostly American - published literature. This chapter identifies injury patterns in UK victims of blast, before establishing a mechanistic hypothesis for the behaviour of the spine in underbody blast.

5.1 Introduction

The previous chapter has shown that the distribution and patterns of blast injury in the spine are poorly understood, but that spinal injury is common in blast and potentially devastating. There is therefore a clear need to improve understanding of the patterns of spinal injury in blast. This section describes a review of the spinal injury patterns in UK blast victims, and starts to develop the relevance of those patterns in understanding the mechanism of injury and in helping to develop better ways to mitigate such injuries in the future. These patterns are compared to those seen in ejection seat injury to explore whether the current practice of using the same injury prediction model for both blast and ejection is valid.

[1] Data from this chapter has been published as "Identifying spinal injury patterns in underbody blast to develop mechanistic hypotheses" in *Spine* [175]

5.2 Method

5.2.1 Patient group

The United Kingdom military maintains a prospectively recorded registry of its injured personnel, the Joint Theatre Trauma Registry (JTTR). This was interrogated by the author to identify all victims with spinal fractures injured in blast incidents between February 2008 and April 2013. The JTTR captured all spinal injuries, including those with a late diagnosis and fatalities. Both mounted and dismounted casualties were identified. Each victim then had their individual JTTR record checked to exclude those who were not victims of IED strikes and who did not in fact have a spinal injury.

Computed tomography (CT) scans were routinely performed on all casualties admitted to hospital. Fatalities were routinely investigated with a CT post-mortem if they did not survive to reach hospital. Each victim had their initial computed tomography (CT) scan reviewed by the author to classify spinal fractures. A consultant military radiologist was consulted in cases of doubt to ensure that all injuries were classified correctly. Results were maintained on a computer database (SPSS, IBM Software).

Basic patient demographics including age, sex, and whether they were mounted at the time of injury were recorded. The presence of a harness, helmet, and body armour was not recorded for security reasons. The Injury Severity Score (ISS) [13], New Injury Severity Score (NISS) [136], and the Trauma Injury and Severity Score (TRISS) [37] were recorded. These scores were extracted from the JTTR database where available, and calculated by the author when not available.

Each fracture was classified by the author according to a standard anatomical and mechanistic classification system. The systems used were selected for simplicity, clarity, and an explanation in their published descriptions of the mechanism of injury. Cervical spine fractures were classified using the Levine and Edwards (C1), Levine (C2) and Ferguson-Allen (C3-C7) systems described in Chapter 3. Thoracic and lumbar zone spine fractures were classified using the McAfee system [121]. The McAfee system was chosen as it allowed a simple, well described mechanistic description for each injury; it is described in Chapter 3. Spinous process fractures were recorded. Open spinal fractures and the presence of foreign bodies were recorded. The abbreviations used to describe fracture patterns are shown in the Table of Abbreviations.

Limb amputations and fractures were recorded using a simple descriptive classification. Pelvic fractures were recorded using the Young and Burgess system [198] with the addition of recording where the pelvis was absent or too disrupted to classify. Skull fractures were recorded as present or absent.

5.2.2 Statistical methods

The database contained continuous data describing age, ISS, NISS, and TRISS. Other data, describing injury classifications, were categorical. Each injury was recorded in SPSS with a code allowing its classification to be recorded. Complex classifications were also re-coded in to groups so all the Magerl descriptions of burst fractures could be analysed together where this improved the quality of analysis. The presence or absence of a fracture at each vertebral level was also recorded.

Parametric data were tested for normality. All fracture classifications were treated as categorical data and analysed using contingency tables and Fisher's exact or Chi^2 tests to compare groups, for example to compare the risk of a fracture at each level between mounted and dismounted victims. A logistic regression analysis was used to identify associations between parametric data and mortality risk.

5.3 Results

5.3.1 Overview and Demographics

The initial search yielded 323 results (Figure 31). 50 had no images available on the system for which we had access consent. 139 of the remainder had no bony injury and were therefore excluded. This left a total of 134 victims, of which 78 were mounted and 46 dismounted (Figure 32). The mean age of each group and mortality risk is shown in Table 10. Dismounted victims had a relative risk for fatality of 2.1 (P=0.000 by Fisher's exact test).

Continuous data were tested for normality using SPSS software's Kolmogorow-Smirnov and QQ tests. Age, ISS, and the number of injured zones in each patient were normally distributed according to the QQ plots, but all other data were significantly skewed. Consequently, parametric data were tested using the Mann-Whitney U test, except for comparing the number of injured zones between groups which was performed with Student's T test.

Figure 33 shows the number of victims with a fracture in each zone, demonstrating that most injuries are thoracic and lumbar. Figure 34 shows the number of victims with a fracture at each vertebral level, divided into survivors and fatalities. Table 11 adds the statistical significance of the differences at each level, and compares the number of fractures and number of involved zones between survivors and fatalities. It can be seen from Table 11 that the number of involved levels does not affect the overall pattern of injury.

The whole study population is divided in to mounted and dismounted subgroups in Table 12. The number of victims with a fracture at each level is clearly demonstrated by Table 11.

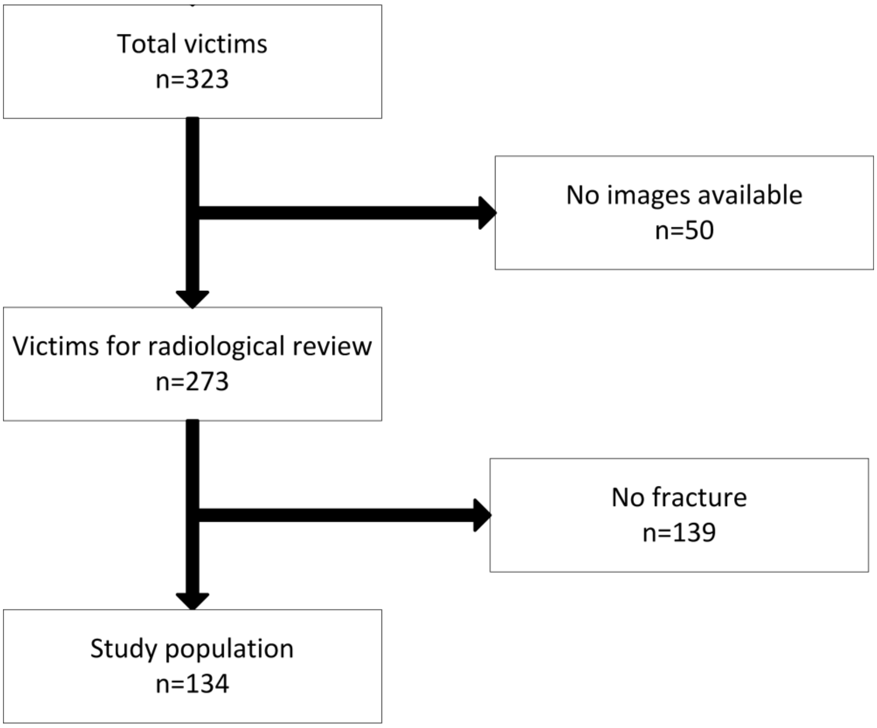

Figure 31: Trauma registry search results.

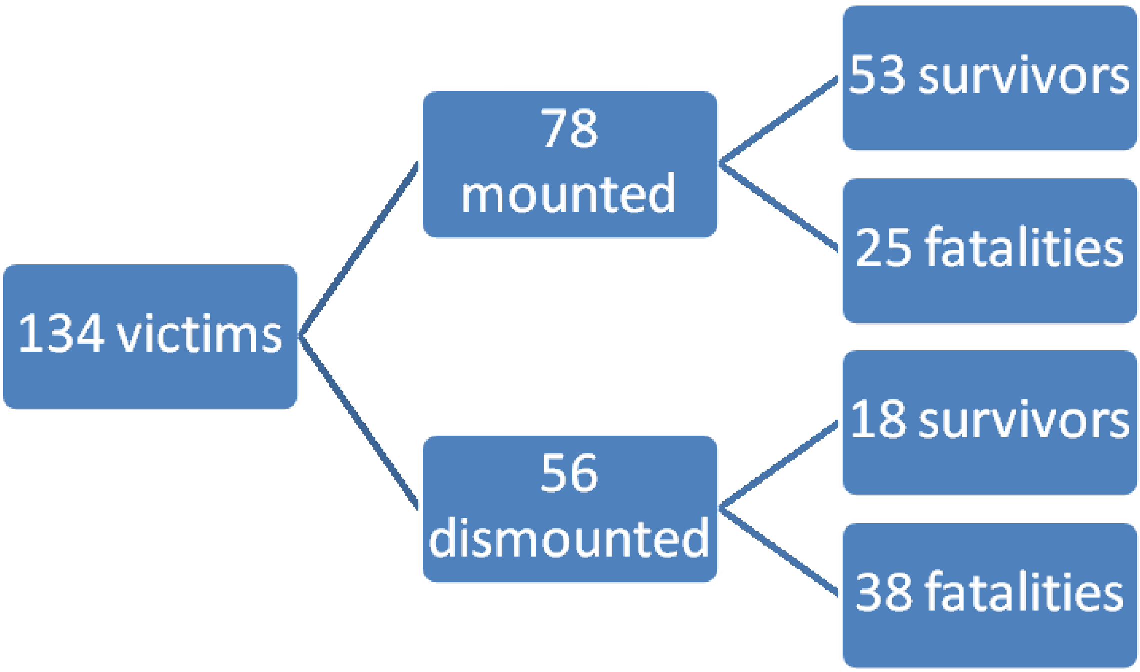

Figure 32: Study population overview showing the number of mounted and dismounted survivors and fatalities.

5.3.2 Dismounted Victims

There were 56 dismounted victims; 18 survived. Of the 56, the mean age was 24.6 years. Each casualty had a mean of 3.36 fractures in a mean 1.34 zones. Table 13 shows the logistic regression analysis for demographic features and injury patterns in dismounted victims. The different types of spinal fracture encountered in each group are shown in Table 14. Dismounted victims are at a higher risk of C1 burst fractures. Mounted victims are at a higher risk of wedge compression and burst fractures. Table 17 shows limb, pelvis, skull, and brain injury risk compared to vertebral fractures at each level.

5.3.3 Mounted Victims

Of the 78 mounted casualties, 53 survived and there were 25 fatalities (13 killed in action, 12 died of wounds later). The mean age of the cohort was 26.8 years (range 18-55). Each casualty had a mean 3.44 fractured vertebrae (range 1 to 21, mode 2, median 2) and a mean of 1.51 involved zones (range 1 to 3, mode 1, median 1). There were 21 cervical, 42 thoracic, and 55 lumbar vertebral fractures. The distribution of injuries is shown in Table 18 and Figure 33~~Figure 33~~.

Of the C1 fractures two were burst fractures, one was a lateral mass fracture, and the 4th an anterior arch fracture. The 6 C2 fractures included four asymmetric pars fractures and two facet dislocations. The subaxial cervical spine fracture patterns are shown in Table 14.

Thoracic and lumbar spine fractures are classified in Table 19. Table 20 shows associated injuries and Table 21 identifies the risk of a spinous process fracture at levels adjacent to a vertebral fracture, which has mechanistic implications that are discussed later. Table 22 shows the logistic regression analysis for mounted victims.

	Number	Mean Age	Survivor	Fatality
Total	134	26	71	63
Mounted	78	27	53	25
Dismounted	56	25	18	38

Table 10: Mean age and mortality in dismounted and mounted groups of UK blast victims.

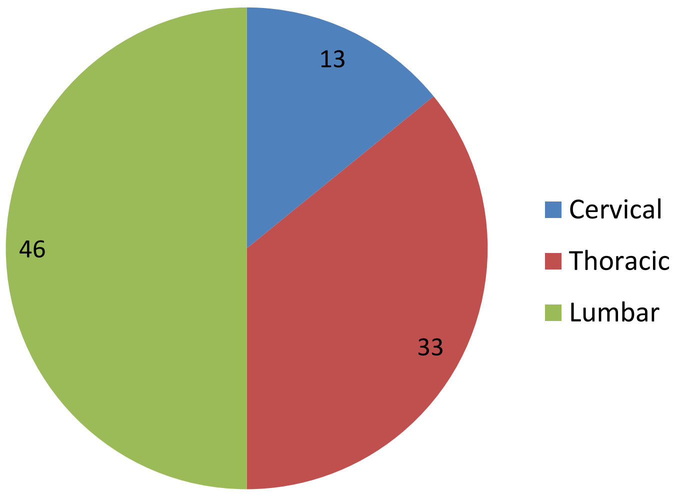

Figure 33: Overall number of victims with a fracture in each zone showing the trend towards lumbar spine injuries.

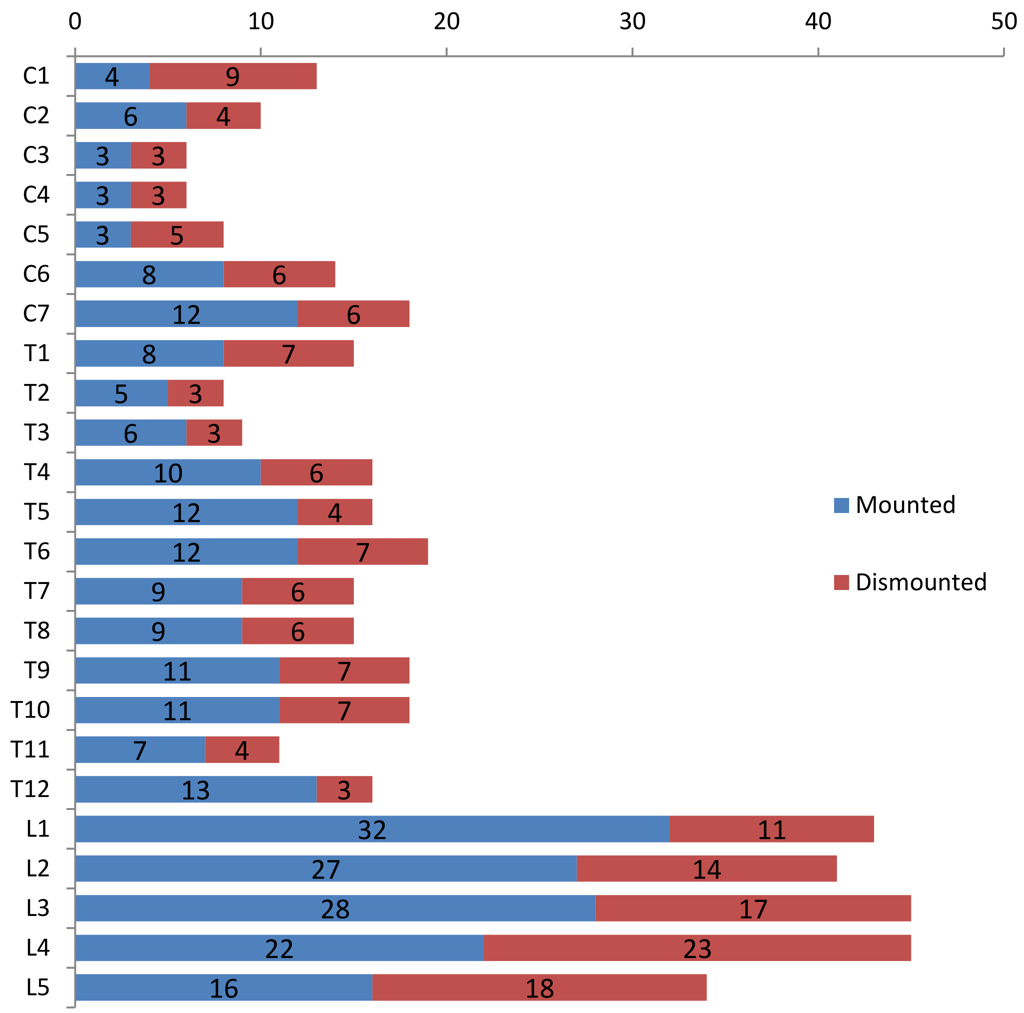

Figure 34: Number of UK blast victims with a vertebral fracture at each level, showing the trends in mounted and dismounted groups.

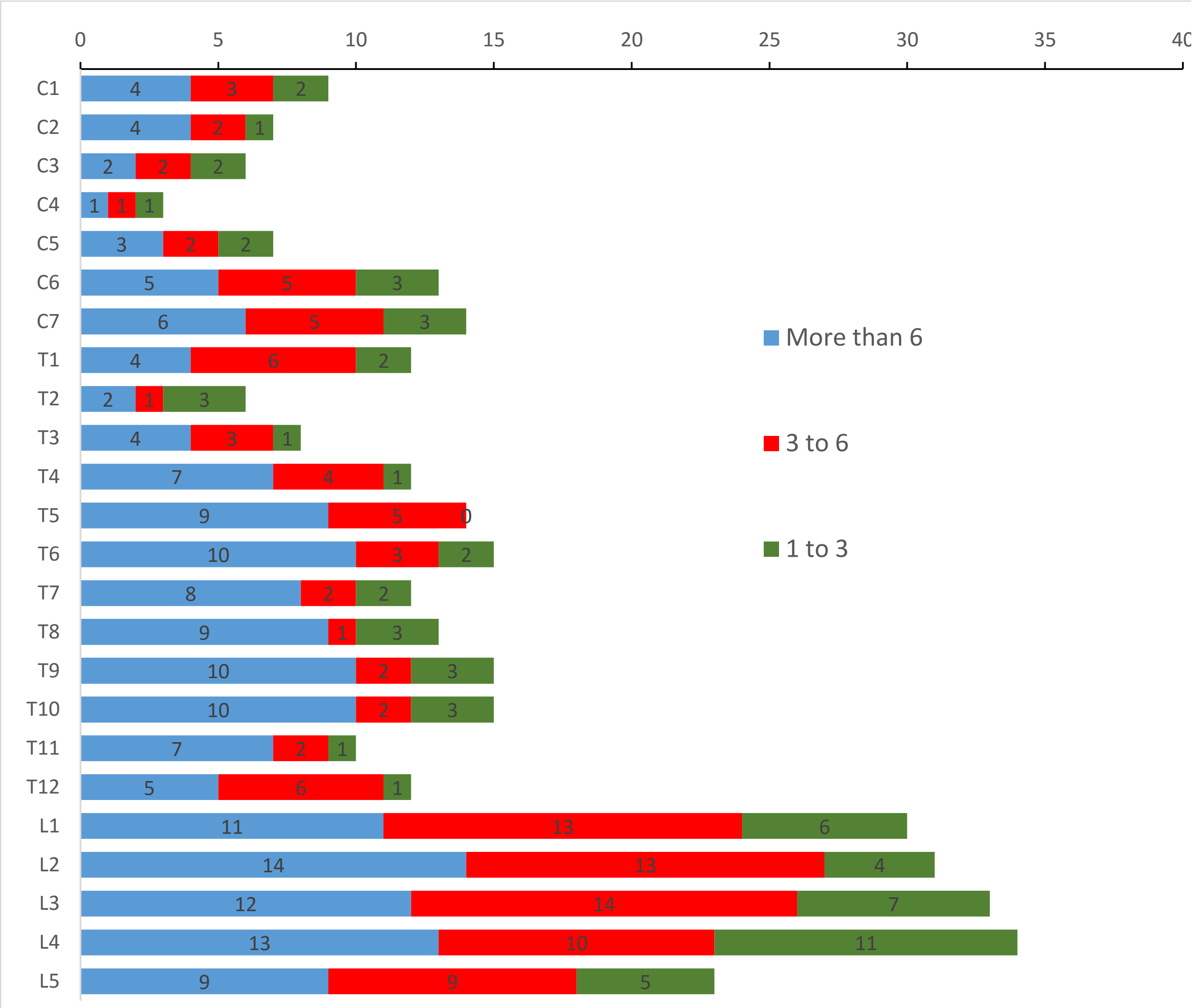

Table 11: The number of fractures in each victim divided in to those with fewer than 3 fractures, those with 3 to 6 fractures, and those with more than 6. This shows that the overall trend of injury pattern is similar however many levels are injured.

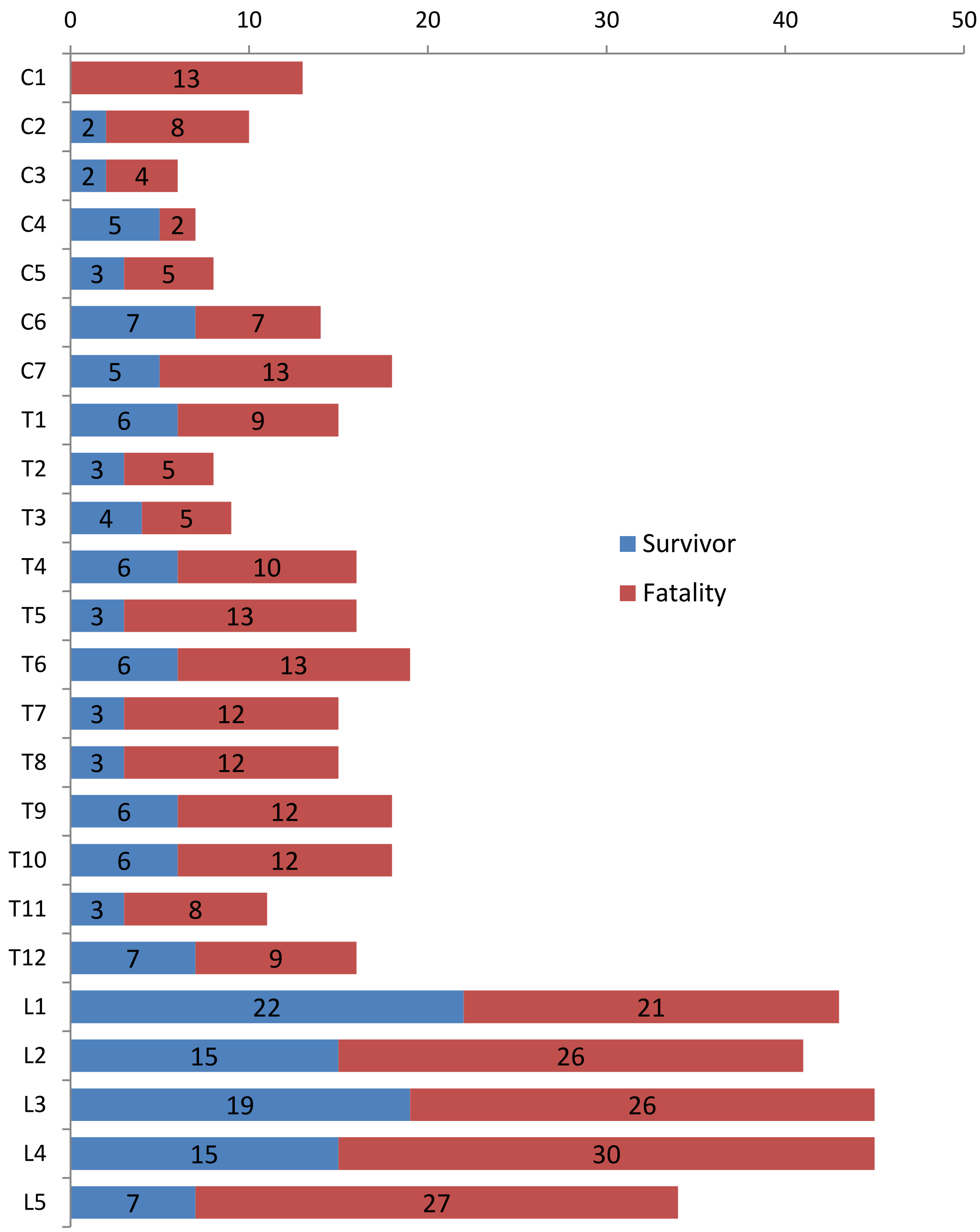

Figure 35: Number of UK victims with a vertebral fracture at each level, showing the different trends between survivors and fatalities, especially in the cervical spine.

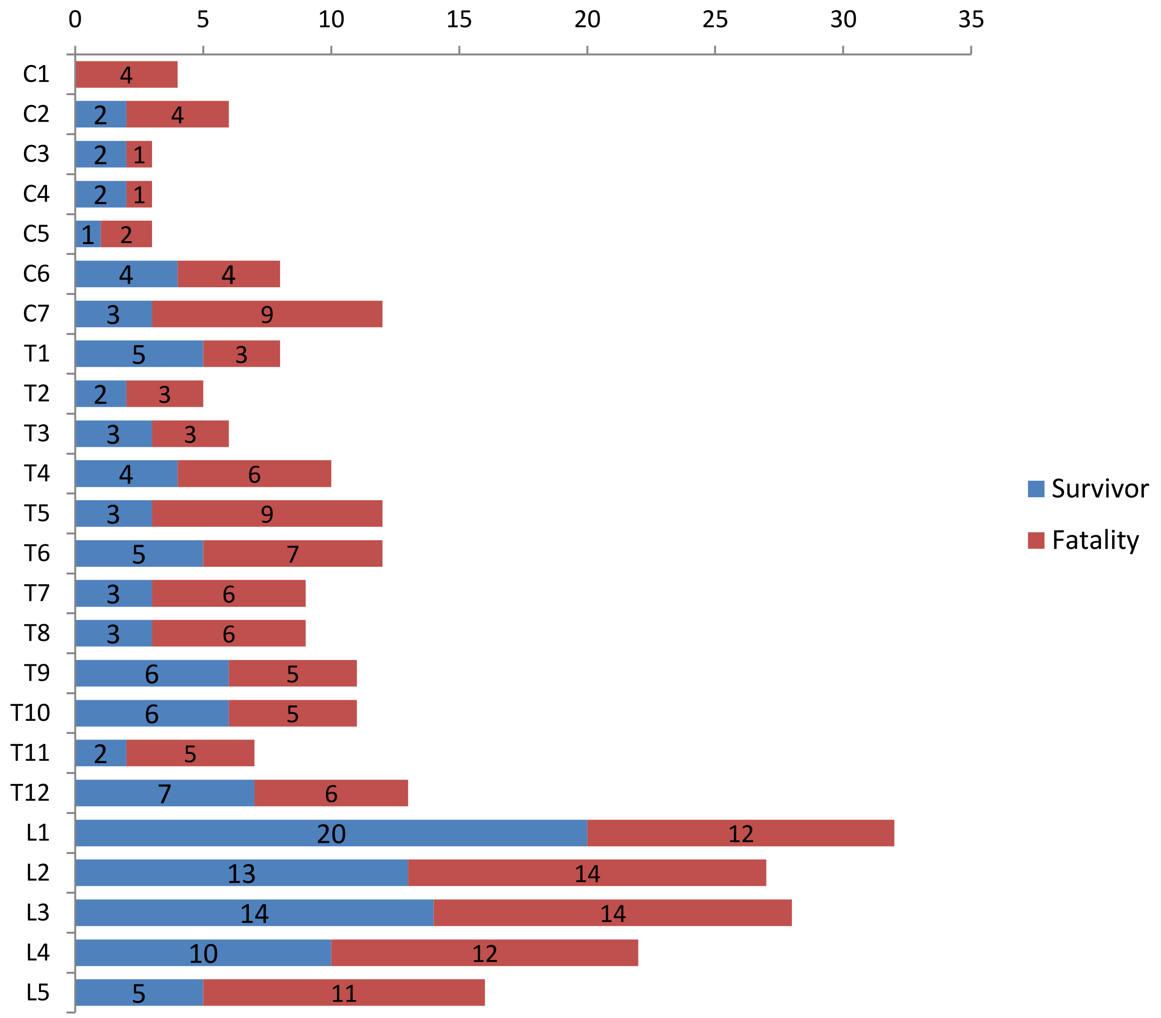

Figure 36: Number of mounted victims with fractures at each level showing survivors and fatalities. Fatalities have more cervical injuries and fewer lumbar fractures.

5.4 Discussion

5.4.1 Overview

This is the most comprehensive review of UK blast victims with spinal injuries yet performed. As the focus of the thesis is on the mechanism of spinal injury following blast in a mounted victim, the patterns seen in mounted and dismounted victims will be discussed separately. However, the main common features in, and differences between, mounted and dismounted victims merit a brief discussion.

The number of levels involved does not appear to affect the overall distribution of injuries. Table 11 shows this clearly. Because the numbers in each group at each level were sometimes small, detailed statistical analysis was not carried out.

There was no significant difference in the ages of mounted and dismounted groups. Overall, the majority of fractures were thoracic and lumbar. Table 12 shows the difference between the risk of fractures at each vertebral level between mounted and dismounted victims. The only statistically significant differences are at C1 and L1. However, Figure 35 shows a slightly different picture in comparing the two groups at each level. In the mounted group, there is a clear trend towards an increased risk of injury at the junctional regions of C7-T1 and T12-L1, where mobile and rigid parts of the spine meet, which is less pronounced in the dismounted group.

The difference in injury distribution is more obvious when considering specific fracture patterns in Table 14. Dismounted victims are at higher risk of C1 burst fractures, and mounted victims are at higher risk of wedge compression and burst thoracolumbar fractures. The lack of a distinct fracture pattern in dismounted victims suggests that the mechanisms of loading and injury are more varied in this group than in mounted victims.

5.4.2 Indicators of Mortality

In mounted victims, mortality risk correlated with ISS, NISS, TRISS, and the number of fractures and zones (Table 22 and Figure 36~~Figure 36~~). In dismounted victims, there was no significant correlation between mortality risk and number of fractures (Table 13).

Figure 34 shows the difference in numbers of survivors and fatalities with an injury at each vertebral level. Strikingly, there were no survivors with C1 fractures; this apparent difference is borne out by the statistically significant difference seen in the risk of C1 and C2 fractures in fatalities in Table 11. It is also clear that C1, C2, and C3 fractures in the dismounted group are strongly associated with skull fractures (Table 16), which may well be the fatal injury in these cases.

There also appears to be a higher risk of mortality with C7 fractures than other levels, mid-thoracic fractures, and L4-L5 injuries. These trends are only true in mounted victims. This suggests either that the mounted victims are subject to different associated injuries, or that these are particularly high-energy injuries.

5.4.3 Dismounted blast

The goal of this thesis is to elucidate a mechanism of spinal injury in underbody blast. Dismounted victims are subject to a broader range of injury mechanisms and variables; the explosion may be any distance from the victim and in any plane including above the victim's head, and it is not possible in many cases to identify their relative positions. Also, dismounted victims may be at greater risk of injury from fragments of environmental debris. Consequently, it may be impossible to derive mechanistic information from dismounted injury patterns. However, a brief discussion of dismounted victims' injury patterns and associated injuries is warranted. Significant indicators of mortality in dismounted victims are seen in Table 14~~Table 14~~.

It can be seen from Table 12 and Figure 35 that there are a few differences in the distribution and pattern of injury between mounted and dismounted victims. The most striking difference, shown clearly by the figure, is that dismounted victims do not have an increased risk of lumbar fractures compared to thoracic fractures, whereas there is a clear trend of an increased risk of a lumbar fracture in the mounted population. However, the only levels where there is a statistically significant difference is at L1 and C7. There is no significant difference in the number of fractures between the two groups.

Table 17 shows the number of victims with a fracture at each vertebral level in the dismounted group. The most common fracture patterns at each level are shown. At most thoracolumbar levels, burst fractures are the most common pattern, suggesting a high energy mechanism of injury.

5.4.4 Associated injuries in dismounted victims

Table 16 shows the limb, pelvis, and skull injuries associated with fractures at each given vertebral level. There is a significant risk of a skull fracture associated with C1-3 and T6 fractures, but no other significant associations were found. The association between skull fracture and cervical fracture suggests that dismounted victims are subject to load through the head and neck. The published data reviewed in this thesis do not give sufficient detail to confirm this notion. One previous study has suggested that dismounted victims are at greater risk of injury than mounted victims, which is not supported by the findings in this series [145]. The lack of associated injuries with given vertebral injuries perhaps suggests a disparate pattern of loading mechanisms in the dismounted group; if all victims were being injured in a similar mechanism by a device exploding at a consistent position in space relative to the victim, one might expect a more consistent pattern of associated injuries. This notion will be considered when discussing mounted injury patterns.

The presence of spinous process fractures throughout the spine with no level at higher risk than any other supports the suggestion that there is not a homogenous loading pattern in these victims. Table 15 demonstrates an increased risk of spinous process fractures in dismounted victims only in the upper cervical spine.

5.4.5 Mounted blast

This thesis is the first single series to determine a mechanistic explanation of the injuries caused by underbody blast through the whole spine. Previous studies have described the demographics and epidemiology of military spinal injuries [17-19, 26, 141, 163, 164, 166-170]. In addition, there have been reports of specific spinal injuries, but no papers have reviewed injury patterns in the entire spine and tried to elucidate a mechanistic explanation for all injuries [24, 25, 27, 63, 81, 95], although some have identified possible mechanisms for individual injuries [63, 108, 149].

5.4.6 Associated Injuries in Mounted Victims

Table 20 shows the associated injuries in mounted victims. Unlike the dismounted cohort, there are several spinal levels associated with a significantly increased risk of a particular associated injury. Lower limb fractures are an exception. Upper limb fractures are more common with C1 and mid-thoracic fractures, either because these are high energy injuries or because, in a sitting posture, much of the upper limb lies at a similar level to these spinal injuries. Pelvic fractures are also associated with mid-thoracic fractures, again supporting the suggestion that mid-thoracic injuries are associated with a high energy trauma. There were no upper limb amputations in the mounted group. Skull fractures were associated with C1 injuries, supporting the possibility that the head is struck by another object during the blast event.

5.4.7 Mechanisms of injury in this series

Cervical spine fractures are less common than thoracic and lumbar fractures in the literature as well as in this series [26, 57]. Burst fractures at C1 suggest an axial load leading to injury [48]. The anterior fracture of C1 suggests an extension-distraction injury and the two bilateral facet dislocation injuries at C2 suggest a combined mechanism of flexion and distraction. There are two possible mechanisms for this. It is possible that the head strikes the inside of the vehicle, or individual equipment is pushed upwards, striking the head. There is an increased risk of a skull fracture with a C1 or C7 fracture, which supports this notion. It is also possible that rather than failing in axial compression the cervical spine is buckling under axial load, leading to the different fracture patterns in the middle of the cervical spine. Nightingale *et al*. noted that increased loading rates lead to increasingly complex buckling dynamics in the cervical spine, along with increasing load at C7 [133]. These fracture patterns also support a buckling mode of failure, and if the head strikes the vehicle it is reasonable to assume that the loads at C1 and C7 would increase, leading to both more fractures at these levels and a combination of C1, C7, and skull fractures; I suggest, therefore, that these data imply a combined head impact and buckling mechanism of cervical spine failure.

Asymmetric fractures suggest either rotation or tilt of the head, which would be expected in a seated vehicle passenger who may be looking out of the window or towards a colleague at the moment of injury. This series showed a small number of asymmetric injuries which may be consistent with such mechanisms.

In the thoracic spine, the majority of fractures occur at T4, T5, and T6 (Table 18). This is the midpoint of the thoracic curve, where the spine is furthest from the centre of gravity of the torso and the line of force of an axial load through the pelvis. Previous series have noted an increased risk of injury in blast at this point [179]. The patterns at this level suggest a mix of flexion and compression, or compressive failure of the anterior spine with distraction failure of the posterior spine. This suggests that the spine is failing under compressive load acting anterior to the vertebral column, along the line of the centre of gravity of the torso or under an axial force up through the pelvis. Spinous process fractures are more common at the levels above and below mid-thoracic vertebral body fractures (Table 21), suggesting tension failures in the posterior spine, which supports this hypothesis.

The thoracolumbar junction is also a point of interest where immobile and mobile parts of the spine meet as the vertebral column loses support from the ribcage at L1. There is a large number of fractures at L1, with an increased number of wedge compression and flexion pattern fractures in this area compared to the adjacent levels. This supports the notion that the lumbar spine flexes as the legs are pushed upwards by the deforming vehicle floor, as suggested by Ragel *et al.* [149]. This is also supported by the large number of pedicle and pars fractures associated with a body fracture at L1 and L2, and the number of lumbar spine spinous process fractures.

Lumbar spine fractures are common in these victims. In the McAfee system, burst fractures are most common; unstable burst fractures suggest a high energy injury. In the Magerl system, burst fractures are further divided in to axial, flexion and pincer burst, where there is a large kyphosis as the vertebra is crushed by those above and below [118]. In the complete burst fractures, there are more flexion than axial burst fractures, suggesting a degree of lumbar spine flexion at the time of injury. In this study the increasing number of burst fractures compared to wedge compression fractures moving down the lumbar spine may be because the vertebral column moves closer to the line of action of the axial force; it may also represent the effect of increasing mass from the body above. The trend towards increasing risk of a pelvic fracture associated with low lumbar fractures also supports this. Lehman *et al*. suggested that lumbar burst fractures occur at lower vertebral levels in military victims than in civilian [108], perhaps because the flexible lumbar spine is supported by body armour, effectively lowering the transition from rigid to mobile spine that otherwise occurs at L1. They also proposed that such injuries may be due to the legs being forced upwards by a deforming vehicle floor combined with the axial load through the pelvis. In this series, burst fractures are common throughout the lumbar spine but are more common at L1 than lower down, which does not support the proposition in Lehman *et al*.

In this series several distinct patterns of injury were noted. All the fractures in the spine were related to axial loading. In the cervical spine there are some extension and distraction pattern injuries suggesting impact with another object, or a buckling mode of failure. In the thoracic spine, the fracture patterns suggest an axial force was applied anterior to the fractured vertebrae, with tension failure of the posterior spinal column confirmed by the presence and location of spinous process fractures. At the junction between thoracic and lumbar spines the fracture patterns suggest the spine is flexed at the moment of injury, possibly because the combination of axial force through the pelvis and upward leg movement causes the pelvis to move forward and in turn the spine to flex. In the lumbar spine the patterns are consistent with high energy axial load through the vertebral body, combined with flexion. In axial acceleration, Vulcan showed that the anterior vertebral body experiences higher load as the thoracolumbar spine flexes [187], suggesting that wedge compression fractures imply a more flexed posture than burst injuries.

Despite the number of injuries being limited, it is possible to identify a consistent pattern of spinal injury in vehicle occupants following an explosion. Each classification system carries a relative uncertainty in assigning mechanisms to the injury, although using more than one system where possible mitigates this somewhat. A blast may occur under any point of the vehicle and the occupant may be sitting in any of a number of locations or postures. Devices may be large or small, with an effect on the load transferred to vehicle and occupant. Each of these may affect the resulting injury pattern. Although it is possible to determine the type of vehicle, seat, harness or personal equipment worn this information is not publishable for obvious security reasons. Nonetheless, in this series, most victims would have been wearing the same standard protective equipment and the same type of harness.

5.5 Summary

This chapter has reviewed the spinal injuries found in UK victims of blast. Dismounted victims of blast have significant injuries and a higher risk of fatality than mounted victims. However, their injuries are disparate with no clear pattern, as would be expected given the wide spectrum of possible causes of injury. In mounted victims, it is possible to see some distinct patterns of spinal and associated injury that allow the mechanism of injury to be considered. A mechanistic explanation of the behaviour of the spinal column during underbody blast has therefore been suggested.

This thesis is intended to support the development of injury prediction models and mitigation systems. The next chapter uses the injury patterns encountered in UK blast victims presented here to consider whether current understanding of the mechanism of injury is adequate to enable these goals, and whether a better injury prediction model is needed.

	Survivor	Fatality	Relative Risk	95% CI RR	p
Mean number of fractures	2.21	4.75			**0.000**
Mean number of zones	1.3	1.6			**0.005**
Cervical	13	26	3.14	1.4-6.9	**0.004**
Thoracic	33	30	1.05	0.5-2.1	0.516
Lumbar	46	45	1.36	0.7-2.8	0.263
C1	0	13	N/A		**0.000**
C2	2	8	5.02	1.02 – 24.60	**0.045**
C3	2	4	2.34	0.41 – 13.23	0.419
C4	5	2	0.55	0.10 – 3.11	0.684
C5	3	5	1.95	0.45 -8.53	0.474
C6	7	7	1.14	0.38 – 3.46	1.000
C7	5	13	3.43	1.15 – 10.26	**0.024**
T1	6	9	1.81	0.61 – 5.40	0.411
T2	3	5	1.95	1.45 – 8.53	0.474
T3	4	5	1.44	0.37 – 5.63	0.734
T4	6	10	2.04	0.70 – 5.99	0.286
T5	3	13	5.90	1.59 – 21.79	**0.006**
T6	6	13	2.82	1.00 – 7.93	**0.05**
T7	3	12	5.33	1.43 – 19.90	**0.011**
T8	3	12	5.33	1.43 – 19.90	**0.011**
T9	6	12	2.55	0.90 – 7.26	0.082
T10	6	12	2.55	0.90 - 7.26	0.082
T11	3	8	3.30	0.84 – 13.02	0.114
T12	7	9	1.52	0.53 – 4.36	0.595
L1	22	21	1.11	0.54 – 2.30	0.853
L2	15	26	2.62	1.23 – 5.61	**0.015**
L3	19	26	1.92	0.93 – 3.98	0.099
L4	15	30	3.39	1.60 – 7.22	**0.002**
L5	7	27	6.89	2.72 – 17.31	**0.000**

Table 12: Number of victims with fracture in each zone and at each vertebral level in each group – mounted and dismounted, showing the different risk in survivors and fatalities. Relative risk=Risk in fatality / risk in survivor. This shows a higher fatality risk with cervical spine fractures and apex of thoracic spine fractures. P value for significance of difference by Fisher's Exact test. CI – confidence interval. R – relative risk.

	Mounted	Dismounted	Relative Risk	95% CI RR	p
Mean number of fractures	3.44	3.36			0.898
Mean number of zones	1.51	1.34			0.124
Cervical	21	18	1.29	0.61 – 7.27	0.565
Thoracic	42	21	.51	0.26 – 1.04	0.079
Lumbar	55	36	0.75	0.36 – 1.57	0.459
C1	4	9	3.54	1.03 – 12.16	**0.042**
C2	6	4	0.92	0.25 – 3.44	1.000
C3	3	3	1.41	0.28 – 7.28	0.694
C4	3	3	1.42	0.28 – 7.28	0.694
C5	3	5	2.45	0.56 – 10.71	0.278
C6	8	6	1.05	0.34 – 3.22	1.000
C7	12	6	0.66	0.23 – 1.89	0.609
T1	8	7	1.25	0.43 – 3.67	0.783
T2	5	3	0.83	0.19 – 3.61	1.000
T3	6	3	0.68	0.16 – 2.84	0.734
T4	10	6	0.82	0.28 – 2.39	0.792
T5	12	4	0.42	0.13 – 1.39	0.183
T6	12	7	0.79	0.29 – 2.14	0.803
T7	9	6	0.92	0.31 – 2.75	1.000
T8	9	6	0.92	0.31 – 2.75	1.000
T9	11	7	0.87	0.32 – 2.41	1.000
T10	11	7	0.87	0.32 – 2.41	1.000
T11	7	4	0.78	0.22 – 2.81	0.761
T12	13	3	0.28	0.08 – 1.05	0.059
L1	32	11	0.35	0.16 – 0.78	**0.009**
L2	27	14	0.63	0.29 – 1.35	0.259
L3	28	17	0.78	0.37 – 1.62	0.580
L4	22	23	1.77	0.86 – 3.67	0.140
L5	16	18	1.84	0.84 – 4.03	0.160

Table 13: Number of fractures, number of involved zones and number of victims with injuries in each zone and at each level, mounted and dismounted groups for both survivors and fatalities. There is a higher risk of L1 fracture in mounted victims, but there is no significant difference at any other level. P value by Fisher's Exact test

	ISS	NISS	TRISS	Number of fractures	Number of zones
Age	**0.288 (0.031)**	**0.310 (0.020)**	-0.165 (0.224)	**0.391 (0.003)**	0.069 (0.614)
Fatality	**0.573 (0.000)**	**0.673 (0.000)**	**0.497 (0.000)**	0.219 (0.104)	**0.381 (0.004)**
ISS				**0.366 (0.006)**	**0.333 (0.012)**
NISS				0.243 (0.071)	**0.379 (0.004)**
TRISS				-0.160 (0.238)	-0.142 (0.297)

Table 14: Logistic regression analysis for dismounted victims comparing ISS, NISS, TRISS, age and number of fractures. (Pearson R, p values by χ^2 in brackets)

	Mounted	Dismounted	Relative Risk Mounted/Dismounted	P value
C1 Lateral Mass	1	3	1.0	0.173
C1 Anterior	1	0		0.397
C1 burst	2	6	0.2	**0.05**
C2 Asymmetric	4	0		0.087
C2 Bilateral	2	2	0.7	0.736
Subaxial CF	3	4	0.5	0.406
Subaxial VC	3	3	1.5	0.778
Subaxial DF	3	0		0.139
Subaxial CE	4	4	1.4	0.682
Subaxial DE	0	3		**0.039**
Subaxial LF	3	0		0.397
McAfee WC	35	7	3.4	**0.001**
McAfee SB	16	4	2.7	**0.047**
McAfee UB	32	11	2.8	**0.018**
McAfee Flexion Distraction	9	3	1.7	0.417
McAfee Translation	7	3	1.4	0.585

Table 15: Number of occurrences of each major type of fracture in mounted and dismounted groups for both survivors and fatalities. Burst and wedge compression fractures are more common in mounted victims, C1 burst fractures are more common in dismounted. Significance of difference by Mann Whitney U test.

	Total	Mounted	Dismounted
C2	15	5	10
C4	10	6	4
C5	8	4	4
C6	5	3	2
C7	7	2	5
T1	13	8	5
T2	18	11	7
T3	13	6	7
T4	8	4	4
T5	8	4	4
T6	14	9	5
T7	16	12	4
T8	20	12	8
T9	14	8	6
T10	13	7	6
T11	17	10	7
T12	10	6	4
L1	16	7	9
L2	17	13	4
L3	44	33	11
L4	42	28	14
L5	44	29	15

Table 16: Spinous process fractures at each level showing the trend towards spinous process fractures at the apex of the thoracic curve and in the lower lumbar spine in the mounted group.

	Lower limb fracture			**Upper limb fracture**			**Lower limb amputation**			**Upper limb amputation**			**Pelvic fracture**			**Brain injury**			**Skull Fracture**		
	n	RR	P	n	RR	P	n	RR	P	n	RR	P	n	RR	P	n	RR	P	n	RR	P
C1	2	0.9	0.547	4	1.2	0.707	3	0.8	0.481	3	1.3	0.338	4	1.0	0.594	4	1.4	0.034	8	1.8	**0.000**
C2	0			1	0.9	0.549	0			2	1.7	0.142	0			3	1.4	0.103	4	1.3	**0.006**
C3	1	1.1	0.615	1	1.0	0.712	2	1.6	0.592	1	1.3	0.452	1	1.0	0.554	1	1.1	0.532	3	1.2	**0.025**
C4	0			1	1.0	0.712	0			1	1.3	0.452	0			0			1	1.2	0.670
C5	1	0.9	0.594	2	1.1	0.593	1	0.6	0.358	1	1.0	0.641	2	1.0	0.569	0			3	1.2	0.158
C6	1	0.9	0.485	2	1.0	0.637	1	0.6	0.200	1	0.9	0.711	2	1.0	0.675	0			4	1.2	0.062
C7	2	1.1	0.654	2	1.0	0.637	3	1.1	0.593	0			2	1.0	0.675	1	1.0	0.532	3	1.1	0.354
T1	3	1.3	0.370	2	0.9	0.513	3	0.9	0.582	1	1.0	0.635	3	1.0	0.582	1	1.0	0.532	3	1.1	0.662
T2	3	1.1	0.662	1	1.0	0.635	1	0.8	0.554	1	1.3	0.452	1	1.0	0.554	0			0		
T3	0			0			1	0.8	0.554	1	1.3	0.452	2	1.1	0.592	0			2	1.1	0.216
T4	0			2	1.0	0.637	2	0.8	0.675	2	1.3	0.289	3	0.2	0.593	1	1.0	0.420	4	0.2	0.062
T5	0			2	1.3	0.611	2	1.1	0.638	2	1.7	0.142	3	1.1	0.328	0			3	1.2	0.079
T6	1	0.8	0.388	4	1.6	0.234	4	1.3	0.693	3	1.5	0.099	4	1.1	0.693	0			5	1.3	**0.022**
T7	1	0.9	0.485	4	2.0	0.172	3	1.1	0.593	3	1.72	0.063	4	1.1	0.401	2	1.1	0.151	3	1.1	0.354
T8	2	1.4	0.654	4	2.0	0.172	2	0.8	0.675	2	1.3	0.289	3	1.1	0.593	0			4	1.2	0.062
T9	2	1.0	0.612	5	2.4	0.084	3	0.9	0.582	2	1.2	0.596	4	1.1	0.693	1	1.0	0.222	4	1.2	0.182
T10	2	1.0	0.612	5	2.4	0.084	3	0.9	0.582	2	1.2	0.596	4	1.1	0.693	1	1.0	0.222	4	1.2	0.182
T11	1	1.0	0.711	2	1.3	0.611	0			2	1.7	0.142	1	0.9	0.615	0			3	1.2	0.079
T12	0			1	1.0	0.712	0			2	2.5	0.079	0			0			2	1.1	0.216
L1	3	1.0	0.619	5	1.2	0.497	5	1.0	0.606	3	1.2	0.393	5	1.0	0.606	2	1.0	0.252	6	1.3	0.071
L2	5	1.2	0.489	6	1.2	0.536	8	1.3	0.375	3	1.1	0.698	9	1.3	0.138	3	1.0	0.103	7	1.4	0.094
L3	6	1.2	0.349	6	1.0	0.606	7	0.9	0.772	4	1.1	0.471	8	1.0	0.589	3	1.0	0.164	6	1.1	0.753
L4	6	1.0	0.585	6	0.8	0.264	11	1.0	0.538	4	1.0	0.614	15	1.7	0.029	5	1.1	0.940	8	1.2	0.569
L5	4	0.9	0.751	7	1.1	0.772	10	1.3	0.399	2	0.9	0.474	12	1.5	0.048	5	1.3	0.940	4	0.8	0.535

Table 17: Associated injuries in dismounted personnel (n=56) for a given vertebral fracture. The only significant association is between skull fracture and cervical or mid thoracic injuries. P value by Fisher's Exact test.

	Total	Survivor	Fatality	Relative Risk (Survivor/Fatality)	p	Most Common Pattern
C1	9	0	9		**0.045**	Burst
C2	4	0	4		0.294	Bilateral facet dislocation
C3	3	0	3		0.544	DE
C4	3	2	1	2.1	0.239	VC
C5	5	2	3	1.1	0.652	CF, VC, CE
C6	6	3	3	1.4	0.374	CE
C7	6	2	4	1.0	0.637	CF
T1	7	1	6	0.7	0.409	FD
T2	3	1	2	1.0	0.696	UB
T3	3	1	2	1.0	0.696	WC
T4	6	2	4	1.0	0.637	TR
T5	4	0	4	0.7	0.294	SP
T6	7	1	6	0.8	0.409	SB
T7	6	0	6		0.162	WC
T8	6	0	6		0.162	WC, UB
T9	7	0	7		0.084	UB
T10	7	0	7		0.084	WC, FD, TR
T11	4	1	3	0.9	0.614	WC
T12	3	0	3		0.544	UB
L1	11	2	9	0.8	0.473	UB
L2	14	2	12	0.7	0.185	SB
L3	17	5	12	0.9	0.515	SB
L4	23	5	18	0.8	0.246	UB
L5	18	2	16	0.7	0.061	UB

Table 18: Dismounted victims (n=56), number of victims with fractures at each level and most common pattern. C1 fractures have a higher mortality risk. See Table of Abbreviations for fracture pattern descriptions.

	Total	Survivor	Fatality	Relative Risk (Survivor/Fatality)	p	Most Common Pattern
Cervical	21	9	12	0.35	**0.006**	
Thoracic	42	26	16	0.8	0.235	
Lumbar	55	35	20	0.8	0.289	
C1	4	0	4		**0.009**	Burst
C2	6	2	4	0.2	0.080	Asymmetric pars
C3	3	2	1	0.9	0.692	CF, VC, LF
C4	3	2	1	0.9	0.692	
C5	3	1	2	0.2	0.239	CE, LF
C6	8	4	4	0.5	0.260	CE
C7	12	3	9	0.2	**0.001**	DF, CE
T1	8	5	3	0.8	0.706	WC, SB, FD
T2	5	2	3	0.3	0.320	
T3	6	3	3	0.5	0.379	WC
T4	10	4	6	0.3	0.067	UB
T5	12	3	9	0.2	**0.001**	WC, UB
T6	12	5	7	0.3	**0.046**	WC
T7	9	3	6	0.2	**0.027**	UB
T8	9	3	6	0.2	0.027	WC, SB
T9	11	6	5	0.6	0.316	WC, SB, FD
T10	11	6	5	0.6	0.316	WC, FD
T11	7	2	5	0.2	0.031	WC
T12	13	7	6	0.6	0.329	WC
L1	32	20	12	0.8	0.463	WC, UB
L2	27	13	14	0.4	**0.010**	WC, UB
L3	28	14	14	0.5	**0.022**	UB
L4	22	10	12	0.4	**0.014**	SB
L5	16	5	11	0.2	**0.001**	UB, SB

Table 19: Mounted victims (n=78), number of victims with fracture at each level, relative risk of fatality and significance of difference by Fisher's Exact test, and most common pattern at each level. Fatality is associated with cervical, mid thoracic and lower lumbar injuries.

	Wedge compression	Stable burst	Unstable burst	Flexion complete burst	Axial complete burst	Pincer Complete burst	Flexion-Distraction	Translation	Spinous process
T1	1	1				1	1		8
T2									11
T3	1								6
T4		1	4	1		3	1		4
T5	3	1	3	3		2	1	1	4
T6	3	2	2	1		1			9
T7	1	1	2			1			12
T8	2	2	1						12
T9	1	1					1		8
T10	1			1			1		7
T11	3								10
T12	4		2	1			1	2	6
L1	7	2	7	2	1	1	1	1	7
L2	4		4	1			2	1	13
L3	3	1	4	1	2				33
L4	1	2	1	1				1	28
L5		2	2	1	1			1	29

Table 20: Thoracolumbar fracture patterns and spinous process fractures - number of fractures at each vertebral level in mounted victims only.

	Lower limb fracture			Upper limb fracture			Lower limb amputation			Upper limb amputation			Pelvic fracture			Brain injury			Skull Fracture		
	n	RR	P	n	RR	P	n	RR	P	n	RR	P	n	RR	P	n	RR	P	n	RR	P
C1	7	1.1	0.787	9	1.4	0.035	5	1.3	0.001	0			10	1.3	0.101	10	2.2	0.682	10	3.6	0.000
C2	14	1.1	0.389	14	1.3	0.064	2	1.0	0.657	0			14	1.1	0.813	8	0.9	0.353	6	0.8	0.560
C3	17	1.0	0.595	11	0.8	0.246	4	1.0	0.536	0			19	1.4	1.597	10	0.7	0.201	10	1.3	0.745
C4	2	1.4	0.583	4	3.9	0.003	2	1.9	0.019	0			4	1.2	0.009	2	1.1	0.301	3	1.3	0.014
C5	2	1.0	0.604	3	1.6	0.151	1	1.1	0.337				4	1.2	0.080	3	1.2	0.572	2	1.1	0.260
C6	1	1.0	0.674	0			1	1.4	0.182	0			1	1.0	0.692	2	1.1	0.522	0		
C7	1	1.0	0.573	1	1.1	0.674	0			0			1	1.0	0.692	0			2	1.2	0.070
T1	1	1.0	0.674	1	1.1	0.573	1	1.4	0.182	0			2	1.1	0.239	1	1.0	0.787	1	1.1	0.426
T2	4	1.4	0.242	3	1.2	0.395	3	1.6	0.007	0			4	1.1	0.260	4	1.3	0.162	3	1.2	0.124
T3	6	1.5	0.172	7	2.0	0.007	3	1.3	0.024				7	1.3	0.046	7	1.6	0.973	7	2.0	0.000
T4	5	2.0	0.098	3	1.2	0.395	0			0			3	1.0	0.706	3	1.1	0.706	2	1.1	0.614
T5	2	1.2	0.491	3	2.0	0.090	0			0			2	1.0	0.653	3	1.2	0.308	3	1.3	0.030
T6	3	1.4	0.365	3	1.6	0.151	0			0			3	1.1	0.379	3	1.2	0.572	2	1.1	0.260
T7	4	1.2	0.487	6	2.0	0.011	0			0			6	1.2	0.067	4	1.3	0.364	4	1.3	0.056
T8	6	1.5	0.172	8	2.5	0.001	1	1.0	0.577	0			9	1.5	0.001	4	1.2	0.198	4	1.3	0.107
T9	7	1.8	0.039	8	2.5	0.001	1	1.0	0.577	0			7	1.3	0.046	4	1.3	0.849	4	1.3	0.107
T10	5	1.6	0.124	6	2.4	0.005	1	1.1	0.468	0			6	1.2	0.027	4	1.4	0.162	4	1.3	0.038
T11	4	1.3	0.445	6	2.4	0.005	1	1.1	0.468	0			5	1.2	0.136	5	1.4	0.194	4	1.3	0.038
T12	5	1.3	0.299	6	1.8	0.021	2	1.2	0.143	0			5	1.1	0.316	3	1.1	0.572	3	1.1	0.380
L1	5	1.3	0.299	6	1.8	0.021	2	1.2	0.143	0			5	1.1	0.316	3	1.1	0.572	3	1.1	0.380
L2	3	1.2	0.670	2	1.1	0.547	1	1.1	0.383	0			2	0.9	0.601	3	1.2	0.079	1	1.0	0.670
L3	3	0.9	0.744	2	0.9	0.503	0			0			4	1.0	0.595	3	1.0	0.079	2	1.0	0.629
L4	2	1.0	0.682	7	1.0	0.791	3	1.1	0.396	0			11	1.1	0.807	5	0.8	0.290	7	1.3	0.234
L5	9	1.1	0.799	8	1.1	0.580	4	1.2	0.046	0			12	1.4	0.126	7	1.1	0.054	5	1.1	0.758

Table 21: Associated injuries with a given vertebral fracture in mounted victims (n=78). Number, relative risk and P value by Fisher's Exact test for victims with injuries at each vertebral level. No significant associations were found.

	2 Above	RR	P	1 Above	RR	P	Same level	RR	P	1 Below	RR	P	2 Below	RR	P
C2													1	1.2	0.003
C4															
C5				1	1.0	0.000	1	1.5	0.000	1	1.0	0.000	1	1.4	0.006
C6	1	1.1	0.018	1	1.1	0.000	1	1.6	0.000	1	1.3	0.252	0		
C7	0			1	1.1	0.000	1	2.0	0.003	0			1	1.3	0.036
T1	1	1.9	0.076	1	1.9	0.149	0			0			0		
T2	0			0			0			0			0		
T3	0			0			0			0			0		
T4	1	1.2	0.279	2	1.5	0.028	2	1.5	0.014	0			1	1.2	0.149
T5	3	1.6	0.003	3	1.6	0.001	1	1.1	0.196	2	1.3	0.009	2	1.3	0.026
T6	1	1.1	0.249	0			1	1.2	0.172	1	1.1	0.249	1	1.1	0.383
T7	0			0			0			0			0		
T8	0			0			0			0			0		
T9	0			0			0			0			0		
T10	0			0			0			0			0		
T11	0			0			0			0			0		
T12	2	1.4	0.020	0			1	1.1	0.249	0			0		
L1	0			2	1.2	0.081	1	1.0	0.452	0			0		
L2	1	1.1	0.280	0			0			0			0		
L3	0			0			0			0			0		
L4	0			0			0			0					
L5	1	1.2	0.182	1	1.2	0.237	0								

Table 22: Mounted victims – vertebral body fracture associated with spinous process fracture risk at levels above and below. Number of victims with both injuries, relative risk and P value by Fisher's exact test. No significant associations were found.

	ISS	NISS	TRISS	Number Fractures	Number Zones
Age	-0.008 (0.946)	0.042 (0.716)	-0.011 (0.925)	0.103 (0.372)	0.103 (0.372)
Fatality	0.859 (0.000)	0.842 (0.000)	-0.729 (0.000)	0.526 (0.000)	0.446 (0.000)
ISS				0.554 (0.000)	0.429 (0.000)
NISS				0.563 (0.000)	0.460 (0.000)
TRISS				-0.412 (0.000)	-0.390 (0.000)

Table 23: Logistic regression analysis for mounted victims comparing ISS, NISS, TRISS, age and number of fractures (Pearson R, p values by χ^2)

6. Is the current injury prediction model suitable? A comparison of blast and ejection injury[2]

As this thesis has discussed, the same injury prediction model is used for both aircraft ejection seats and underbody blast. In order to understand whether this is appropriate, the principles of aircraft ejection seats are reviewed in this chapter, and the injury distributions in ejection compared with those seen in blast in order to assess from scratch the validity of using the same model for both scenarios.

This chapter aims to identify whether or not the DRI model is clinically valid for predicting spinal injury in under-body blast attacks against vehicles. The validity of a model like DRI depends on the behaviour of the model *in vitro* matching the behaviour of the spine *in vivo* [147]. Traditional methods of validating a model depend on demonstrating that its measured behaviour is identical to that of living specimens or post-mortem experimental models. However, the behaviour of the spine is complex, and the behaviour of all its elements at the high loading rates seen in blast is not well understood. While it was recognised as long ago as 1974 that lumped-parameter models such as DRI are imperfect, improved models have not been developed [147].

6.1 Aircraft Ejection Seats

The development of high performance aircraft during World War II necessitated improved means of allowing the crew to escape a damaged aircraft as force required to escape from an aircraft, and the speed required to do so in time, became impossible. Work in Germany and Sweden lead to the concept of a seat fired vertically from the aircraft by an explosive charge [144]. Almost all modern fast jet aircraft are now fitted with an ejector seat such as those seen in Figure 37 and Figure 38, and to date seats made by Martin Baker have saved 7,426 lives [2].

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[2] Data from this chapter has been published as "Blast injury in the spine: Dynamic Response Index is Not an Appropriate Model for Predicting Injury" in *Clinical Orthopaedics and Related Research.* [176]

The Martin Baker aircraft company, based in Denham, was responsible for the development of British ejector seats and also for early attempts to predict the injury patterns caused by ejection. In 1945 a test rig was constructed allowing a 200 lb dummy to be fired to a height of 10 feet using an explosive charge [144]. The first human subject was Bernard Lynch, a fitter at the company, who was initially fired up the rig to a height of 4 feet 8 inches. Subsequent tests with larger charges launched him to 10 feet at which point he reported back pain. The second subject was Charles Andrews, a journalist from *Aeroplane* magazine. He was also fired to 10 feet, but suffered a vertebral fracture in doing so. Consequently, the seat designer sought the advice of an orthopaedic surgeon and attempted to establish a set of criteria to reduce the risk of vertebral injury during ejection.

As has been shown in this thesis, victims of underbody blast in a vehicle are exposed to high acceleration over a short time period in a mainly upwards direction. This mechanism is, at face value, not dissimilar to that seen in an ejector seat. This chapter reviews the principles of ejector seats and the existing research and models used to minimise ejector seat injury and the injury patterns caused by aircraft ejection reported in the published literature. The lessons learned in ejector seat design that can be drawn across to blast injury are discussed.

6.1.1 Ejection Seat Principles

Ejection seats have been developed by several companies worldwide, but the principles are similar for most types of escape system. Ejection is initiated by the pilot pulling a seat handle [113]. The phases of ejection can be summarised as [113]:

1. Escape pathway clearance;
2. Seat preparation;
3. Ejection gun firing and rocket burn;
4. Windblast;
5. Drogue parachute deployment;
6. Main parachute canopy deployment;
7. Parachute landing;

Pulling the initiation handle on the seat fires an initiation cartridge which pipes gas around the seat to control the remaining phases of ejection [113]. The first part of the ejection sequence is the clearance of the pathway to ejection. Most modern fast jet aircraft use a miniature detonating cord (MDC) in the canopy to shatter it prior to ejector seat activation. Alternative systems include canopy separation and canopy penetration by sharp penetrators affixed to the top of the seat, although these have been associated with an increased risk of injuries [191]. Modern seats incorporate a powered harness retractor designed to improve the pilot's posture during ejection [113]. Leg restraints are also fitted and some seats incorporate arm restraints. These are designed to place the pilot in an upright posture, intended to reduce the incidence of spinal injury, and to stabilise the limbs and reduce the incidence of flail injuries.

The main ejection gun then fires. This is the phase of ejection which warrants comparison with underbody blast injury. An explosive cartridge accelerates the seat vertically up guide rails. This is associated with a significant axial load to the spinal column and is felt to be the main cause of vertebral compression fractures in aircraft ejection [11, 113]. A time-acceleration curve from a ballistic seat is shown at Figure 39. “Ballistic” seats are propelled purely by an explosive charge. Many modern systems use rocket packs to produce greater velocity and lower initial acceleration, but purely ballistic seats remain in use especially where light weight is important [2]

Most seats initiate drogue parachute deployment by a cable fixed to the aircraft, and then initiate the main parachute depending on speed and altitude after a time delay using a clockwork mechanism. The earliest seats depended on the pilot to activate each stage and separate the seat and parachute. The latest seats are electronically controlled [2].

The later phases of ejection provide confounding sources of spinal injury. On leaving the aircraft the pilot is immediately subjected to high speed wind blast with a risk of limb flail injuries. Windblast itself has produced fatal injuries in some high speed ejections [171]. The drogue parachute, intended to stabilise and slow the seat, induces a shock load in the spinal column which may lead to injury. The main parachute then opens with significant shock loads, and the possibility of the upper parachute harness striking the helmet and injuring the cervical spine. This impact has been shown to cause helmet damage [11] and may be a source of cervical spine injury. Finally, on landing, the pilot may strike the ground at a significant velocity with a risk of lower limb and spinal fractures [113].

There are therefore several stages during the ejection sequence which could lead to spinal injury. The ejection gun produces a force similar in direction to the force experienced in underbody blast and will therefore be considered in more detail.

6.1.2 Forces during ejection acceleration

The first ejector seats in service produced peak accelerations of approximately 12g with an onset rate of 1100g s^{-1} [11]. This was associated with vertebral fractures before the end of the War. It was recognised early that the "jolt", or rate of rise of acceleration, was critical as too fast a rate of acceleration means that the spinal muscles do not reflexively support the spinal column fast enough; this takes about 150 ms [11, 74]. The limits of jolt and acceleration were therefore set initially at 300g s^{-1} and 25g, respectively.

Early ballistic seats used a single cartridge to accelerate the seat at 16 to 18 ms^{-1} [11], later increased to 24 ms^{-1}, to improve the ability to escape the aircraft at high speed and low altitude. This change was noted to increase the incidence of vertebral injury [65] from 10 to 35%. This was the trigger to introduce rocket assisted seats, in which a rocket fires shortly after the initial ejection gun. This system allows the pilot to be propelled further from the aircraft, therefore allowing a much greater safe ejection envelope, and at the same time reduces the velocity needed from the ejection gun [11]. Rocket assisted ejection seats are therefore also less similar to blast injury than ballistic seats.

Figure 37: Martin Baker Mk 10 ejection seat. Note the white rocket packs under the seat. Taken from [134]

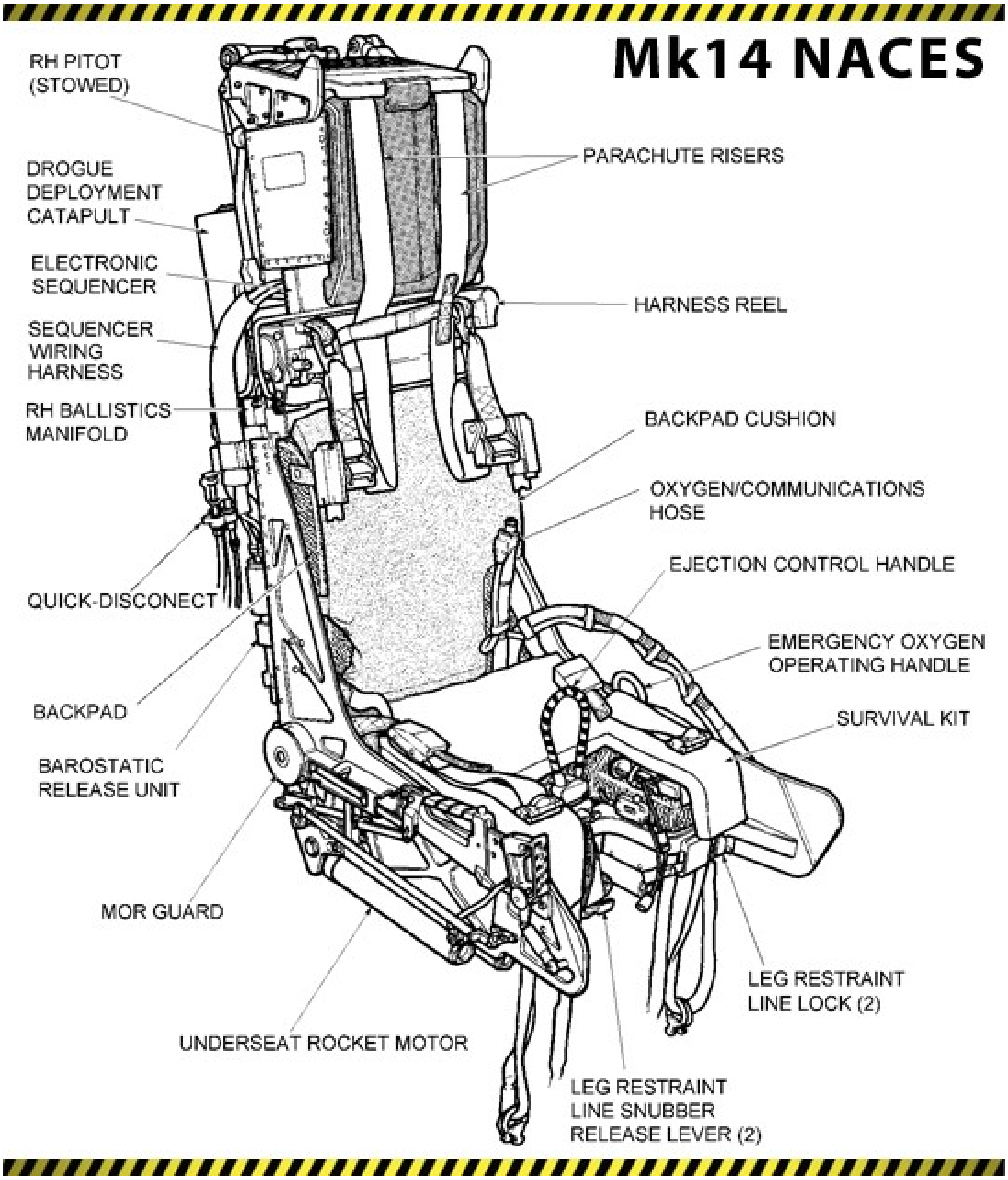

Figure 38: Martin Baker Mk 14 ejection seat, showing the controls and main features. Taken from Martin Baker website, with permission [2]

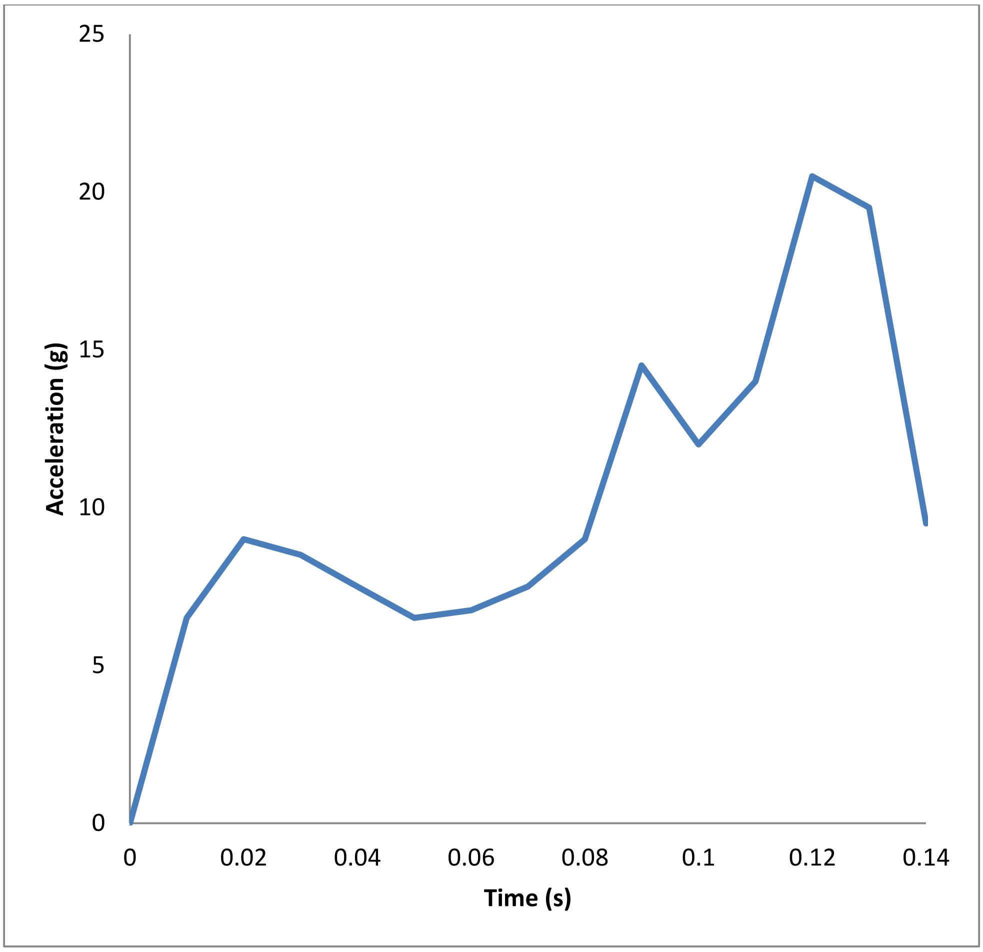

Figure 39: Acceleration - time curve for early ballistic seat showing peak acceleration of 22g at 0.12 seconds. Adapted from Stewart *et al.* [182].

Comparing two standard Martin Baker seats demonstrates the difference between rocket assisted and ballistic seats [12]. The Mk 6 rocket assisted seat produces a peak acceleration of 15g, with a rate of onset of 200g s^{-1} over approximately 0.45 seconds and achieves a velocity of 80 fts^{-1}. A late model ballistic seat, the Mk 4, also produces a peak velocity of 80 fts^{-1} but accelerates at 20g, with a rate of onset of 240g s^{-1}, and accelerates over 0.21 seconds.

The effect of axial acceleration on the load borne by each vertebral level is clear when the effect of the pilot's mass is considered. Approximately 7% of body weight is applied to C4 in normal conditions, increasing to 50% at L1 and 60% at L5 [160]. At 1g this equates to a load of 42kg at L5 for a 70kg pilot. At 20g, this rises to 840kg. The breaking load of a lumbar vertebra ranges from 520 to 635 kg in static conditions [143].

6.1.3 Spinal alignment during ejection

The posture of the pilot in an ejector seat was felt to be critical to minimising injury risk early in their development as it was believed that the spine was at lower risk of injury if aligned along the axis of ejection acceleration, even before the behaviour of the spine in ejection was understood [12, 160].

Consequently, the first ejection seats were fired by a handle placed above the pilot's head which, when pulled, placed a screen in front of the face to hold the head in position and prevent injury from fragments of canopy and other objects. However, it became clear that such designs required more time to initiate ejection than was available at low level and high speed [11] and later seats therefore used a firing handle mounted on the seat pan. In order to compensate for the loss of posture control, these seats began to incorporate powered harnesses which pulled the pilot's shoulders back in to the best possible position during seat initiation. This led to a reduction in the incidence of vertebral fractures in US Navy ejections [76].

The effect of early seat design on vertebral fractures was reviewed by Levy [112]. This paper discussed the effect on vertebral fractures of different seat back and head-box designs, with regard to the flexion-extension alignment and curvature of the spinal column. The trunk-thigh angle was also considered; the lumbar spine lordosis is determined by hip flexion and excessive flexion, beyond 135°, causes the pelvis to rotate forwards, eliminating the lordotic curve.

The position of the head during the ejection sequence is important. Cervical spine fractures have been associated with ejection in two-seat aircraft with a significant distance between the canopy and pilot, as the pilot's head invariably flexes forward during ejection [41] so may strike the canopy in a flexed posture; this has been associated with unstable cervical spine fractures [11]. As the head flexes forward, the cervical and thoracic spines flex [160]. This leads to an increased distance between the mass of the head and thorax, and the apex of the thoracic curve. With the force of the ejection gun from below and the mass of the torso acting from above there is therefore a significant force at the apex of the curve. This is both the area of maximum kyphosis, and therefore maximum anterior vertebral body force, and the area with the largest moment [82, 160]. Even with a tight harness, photographs of pilots during ejection [41] have shown the head flexed almost to the point of contact with the thigh. Conversely, Shannon [171] reported one case of ejection with a loose harness in which the pilot reported his chest having contacted his thigh during ejection, but had no spinal injury.

6.2 Ejection Seat Injury Patterns

The published literature describing ejection seat injuries was reviewed by the author. A Pubmed and Google Scholar search using the terms “ejection/ejector seat” and “injury” was used to trawl for papers which were then reviewed to identify spinal injuries. There have been several reviews of the injury patterns seen in aircraft ejection. Those of most interest to this thesis focus on ballistic seats, which in the UK were produced until 1966 with the introduction of the Martin Baker Mk 6 [2]. Soviet ballistic seats were produced in a similar era. Not all of the published papers separate seat types, or give great detail in the injuries seen.

6.2.1 Ballistic seats

Fryer [65] produced an early review of Royal Air Force experience with ejection seats. This review is of note because it predates the introduction of rocket-assisted seats. The majority of spinal fractures were thoracolumbar. 25 of 41 ejectees had multiple fractures. It was also noted that increasing age was associated with a higher risk of fracture. Fracture patterns, however, were not recorded.

Jones *et al.* [93] described early American experience with Martin Baker seats in the US Navy. The reported spinal injury rate was 21% in ballistic seats. Most of the fractures were between T8 and T11. The paper reports that most of these ejectees were able to return to full flying duty.

The distribution of fractures from all ejections using ballistic seats is shown at Figure 40.

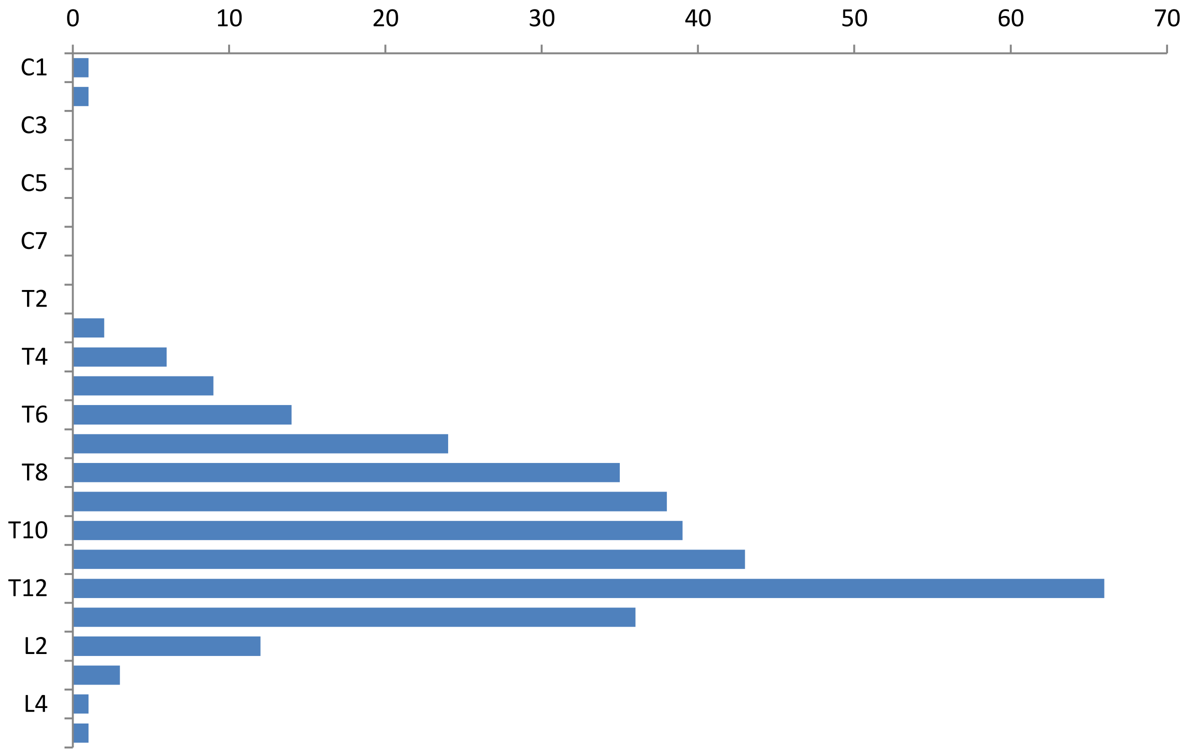

Figure 40: Number of fractures per vertebral level from papers reporting injuries relating to ballistic seats . Most fractures occur at T12 with a decreasing number in the lower lumbar spine.[65, 93, 124, 132].

6.2.2 Rocket assisted seats and mixed case series

The British experience with ballistic and rocket assisted seats was reported by Anton [11]. This paper gave a broad overview of the background of seat development and the inception of injury tolerance models. The reported vertebral fracture incidence was reported to be between 29% (Mark 9 rocket seat) and 69% (Mark 2 ballistic seat). Cervical spine fractures were reported to be extremely rare and were not felt to have been caused by the ejection gun acceleration. The majority of vertebral fractures occurred at T12 and L1, with lumbar fractures more common in the earlier seat designs.

Lewis [114] reported a series of 232 rocket assisted ejections in Royal Air Force aircraft, in which 29.4% of ejectees sustained spinal fractures. The paper divided injury patterns according to seat type. Most fractures were clustered at T12/L1 and T5 to T8, with a higher incidence of mid thoracic fractures in the Tornado aircraft. All but two fractures were wedge compression with only two burst-pattern fractures noted. Pilot stature and leg and arm measurements did not affect the risk of fracture.

The Italian experience was reported by Rotondo [160]. Of 100 ejections, 15 had spinal fractures. T12 and L1 were the most commonly injured vertebrae. It was noted that ejections from the Starfighter equipped with a Martin Baker seat produced thoracolumbar fractures, whereas the same aircraft fitted with a Lockheed seat produced mid-thoracic fractures. This may be because the Lockheed seat produces a faster rate of G onset than the Martin Baker seat. This is comparable with Lewis's paper [114] in which Tornado ejections produce more mid-thoracic injuries than those from similar seats in other aircraft, possibly because the Tornado ejection gun produces a higher acceleration than that seen in otherwise identical seats in different aircraft types. A short case series of ejections from Hawk aircraft with Martin Baker rocket assisted seats [89] demonstrated thoracic spine wedge compression fractures in two out of three patients.

A case series of Bulgarian ejections [124] described injuries sustained by 60 ejectees in both ballistic and rocket assisted Soviet era seats. The paper separated each patient, so it was possible to extract the injury patterns in ballistic seats. The overall incidence of spinal fractures was 16%, but was only 8% in ballistic seat ejections. However, the ballistic ejections reported were often out of envelope and the fatality rate was 24%, with no details of the injuries sustained by fatalities reported.

The German experience of rocket assisted seats was published by Werner [189]. 16 ejectees were reported with 21 vertebral fractures. Two fractures of C2 were noted, both in pilots with their heads flexed forward at the moment of ejection. L1 was the most commonly injured vertebra. There was one intervertebral disc injury.

The Finnish Air Force reported a mix of rocket and ballistic ejections using both Martin Baker and Soviet seats [186]. The reported incidence of spinal compression fractures was 18%. The paper states that the fractures were thoracic but does not give further detail.

The Australian experience of ejection injury was reported by Newman [132], covering all ejection seats from 1951 to 1992. The overall incidence of vertebral fractures was 35%. In the ballistic seats, 56 vertebral fractures were reported among 67 ejectees with the majority of the fractures between T8 and T12.

American ejectees from the Gulf conflict of 1991 were reported by Osborne and Cook [135]. All these ejectees were in rocket seats. The overall incidence of vertebral fractures was 33%. The most common injuries were thoracic or cervical compression fractures but there were lumbar transverse and spinous process fractures in two patients.

Bowman [31] produced a review paper covering all ejection systems. He suggested that cervical spine fractures are less common than thoracic and lumbar injuries, and are less likely to be caused by the acceleration of ejection than by impact with the canopy or parachute harness interference. This suggestion is supported by Anton in the UK [11]. Bowman also stressed the importance of correct pre-ejection positioning in reducing the injury rate.

Auffret [12] produced a review paper of worldwide ejection incidents. Overall vertebral fracture rates in survivors of ejection ranged from 10 to 47%. Multiple fractures occurred in 40.8% of pilots with vertebral fractures. The paper commented that the fractures in pilots with multiple fractures were less severe than those with a single level injury, but did not detail the nature of those fractures. It was also noted that parachute landing injury reported in paratroops produced a distinctly different pattern of fractures, suggesting that the fractures in ejection injury are not largely caused by the force of landing. This paper also highlighted the importance of correct seat position in preventing excessive thoracic kyphosis. The role of the head nodding forward under acceleration, with a point of flexion between the mid-thoracic and thoracolumbar spine, was emphasised.

The distribution of fractures in all ejections from both ballistic and rocket seats is shown at Figure 41. This is based on all the published literature found; no duplicate publications were identified.

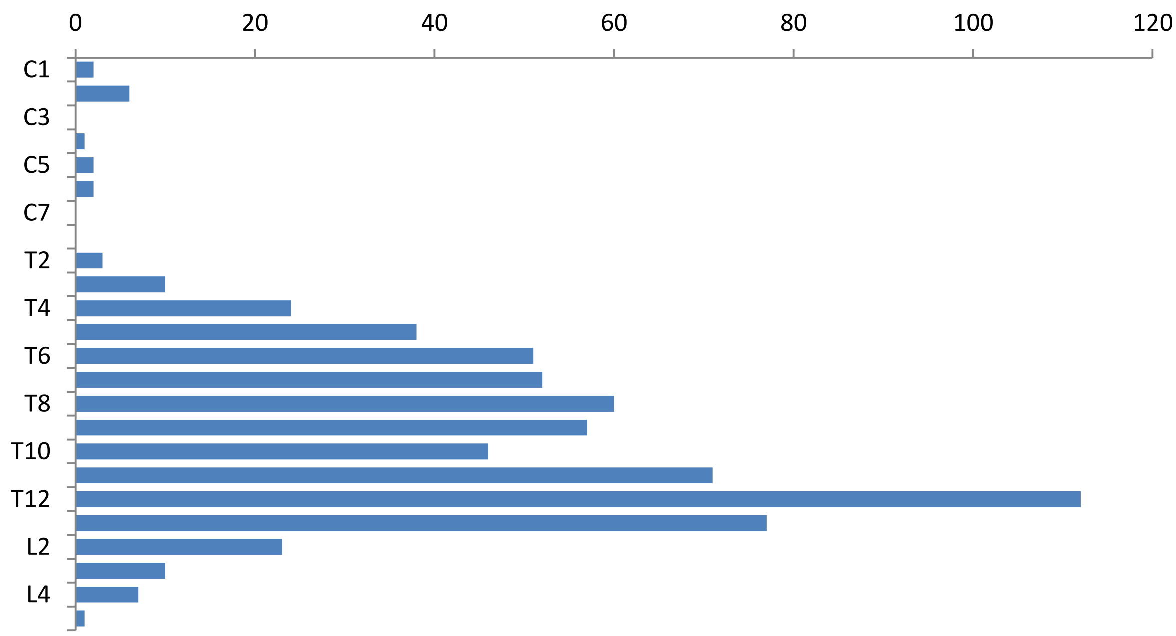

Figure 41: Number of fractures per vertebral level in the published literature, all seat types. The trend is similar to that seen in ballistic seats [11, 12, 31, 42, 65, 77, 89, 93, 99, 114, 117, 124, 132, 135, 160, 186, 189].

6.2.3 Fatalities during ejection

Fatalities during ejection were reported by Lowry *et al.* [117] in 1994. This paper covered both ballistic and rocket-assisted ejection seats. 57 post-mortems were reported and the events surrounding ejection were reviewed. The majority of fatalities were associated with ejection outside the design limitations of the seat, such as at too low an altitude for an older seat design or close to inverted. There were several cervical spine fractures which were felt to have been caused by windblast, not related to the acceleration of initial ejection.

6.2.4 Seat cushion effect

Ejection seat propulsion systems are designed to produce a tolerable load on the spinal column. Attention has been paid to the interface between seat and pilot to control the load transmitted to the spine. A simple model [12] describes the seat-pilot system as a pair of masses connected by the elasticity and damping capacity of the cushion. A rigid seat cushion produces identical acceleration for the seat and pilot. A soft cushion transmits the acceleration to the pilot with a phase delay and over a longer duration, once the cushion has been compressed fully. However, as the seat cushion starts to compress, the seat is accelerating, so the effective point of impact occurs at a point when the seat is moving faster than it would be with a rigid cushion in place.

Auffret *et al.* [12] report one example of injury caused by a padded seat cushion. A pilot ejected in controlled flight from a damaged aircraft at an ideal altitude and speed and with careful attention to his harness and posture. He had modified his seat with a comfortable cushion. At the moment of ejection, he felt severe back pain. On landing his parachute snagged in a tree so there was no impact with the ground. He was noted to have a fracture of T8 vertebra, which would not have been expected with the ejection seat used.

Similar principles of the interplay between seat and occupant apply to crashworthy seats for aircraft and blast resistant seats for vehicles. Either the seat mounting system or the cushion seek to absorb energy during impact loading, reducing the peak acceleration. Energy absorbing seat cushions have been shown to reduce the load experienced by the lumbar spine in underbody blast tests [9]. Crashworthy seat designs aim to reduce the peak load seen by the lumbar spine, using a system of dampers or energy-absorbing rails and a sliding seat to moderate energy transmission. They have been shown to be effective in reducing the predicted injury rate in a cadaver model [43].

It is suggested therefore that the connection between seat and occupant is likely to have an effect on the risk of injury during underbody blast. However, the interaction between seat and occupant depends partly on the behaviour of the spine, which is not well understood.

6.3 Dynamic Response Index: is a new injury prediction model needed?

DRI is a simple, lumped parameter model of the spine during axial acceleration. As discussed in Chapter 2 it does not consider how the spine moves during blast. As discussed previously in this thesis, vehicle designers use several strategies to try to reduce the risk of spinal injury following underbody blast. These include vehicle body modifications such as the V shaped hull, and energy absorbing seats. The tests for these are mandated by NATO and rely on the behaviour of a dummy placed in a vehicle seat and use the DRI to equate the dummy's behaviour with a risk of spinal injury to the vehicle occupant. However, DRI was not intended to be used for blast tests and has not been validated in blast.

If DRI is to be considered adequate for both ejection injury and blast tests, it would be necessary for the mechanisms of injury in ejection and blast to be similar. Although the direction of loading is apparently similar in these two scenarios, the rate of loading may be significantly different. It is therefore necessary to compare the mechanisms of injury in blast and ejection to assess the validity of DRI in both situations. Since the mechanism of injury in blast and ejection is not fully understood, a secondary means of testing the validity of DRI is appropriate. If blast and ejection have similar mechanisms, they would be expected to have similar injury patterns and if this is the case then DRI would be valid in both cases. However, if the patterns of injury are different, then the mechanism of injury must be different, and DRI would not then be suitable for both types of loading.

This section therefore compares the patterns of injury in the UK blast spine cohort with those in the published literature in order to validate or refute DRI as a model for both scenarios.

The distribution of spinal injury in UK victims of blast was identified in Chapter 5. Using these data, a pairwise analysis was performed to compare the risk of injury at each level between ejection and blast with Fisher's exact test used to measure significance. Four papers were identified with sufficient detail for the ejection cohort of this study. They reported 258 fractures in 189 patients [65, 114, 124, 132]. These papers did not classify the fractures anatomically or mechanistically. A significance level of $p = 0.05$ was set for this study.

Table 24 compares the risk of a fracture at a given zone or level in aircraft ejection with the mounted group of victims in this study. The percentage of injuries at each level is shown in Figure 42.

6.4 Discussion

There has been some concern over the validity of DRI in aircraft ejection. DRI under-predicts the spinal injury rate, especially in later ejection seats [11], possibly because the later seats have multiple ejection guns and rocket assistance so do not provide a single input pulse. Table 23 is taken from Anton's paper and compares the DRI prediction of fracture incidence with the observed incidence from UK ejections. The data on which DRI is based are approximate, especially with regard to the failure strain of vertebrae under high loading rates and the complexity of the vertebral column as a system of springs and dampers. The model also only considers motion in the Z axis (vertical / axial), although as the spine flexes during axial acceleration, there is also movement in the X and Y axes [49].

Seat Type	DRI predicted fracture rate	Observed fracture rate
Mk 2	80-100%	69%
Mk 3	80-100%	65%
Mk 4	40%	39%
Mk 6	4%	65%
Mk 7	4%	50%
Mk 9	4%	29%

Table 24: Predicted fracture incidence for different types of ejector seat, using DRI model, compared with fracture rate seen in reality; there is a marked difference with modern rocket assisted seats [11].

In reality, therefore, the DRI model may be over-simplified for complex loading scenarios. DRI models the spine as a single spring, mass and damper. The spinal column may be considered to be a connected series of masses, springs, and dampers [12]. The masses are the head, thorax, arms, abdomen, pelvis, and lower limbs; the springs are intervertebral discs, muscles, ligaments, and tendons. As the spine is accelerated from below, each level of intervertebral disc is compressed, and each vertebra moves vertically with respect to its neighbour. The spinal column also flexes as it is compressed under axial load. However, the behaviour of the column as a whole during the event of high-rate axial acceleration is not well understood.

The DRI model may be as valid in blast as it is in ejection if the patterns of injury in each group are demonstrably similar. It is clear from Figure 42 that the distribution of injuries is different. In this study, the ejection injury group had more thoracic than lumbar or cervical spine injuries. The blast group had a higher incidence of lumbar fractures. This analysis (Table 24) shows that there is a significantly different risk of fracture at all levels of the lumbar spine and at all but one level of the cervical spine. There was not a significantly different risk of fracture in the lower thoracic spine, where the majority of ejection seat victims were injured. Therefore, although there is some similarity in the injury pattern, the large difference in risk at the cervical and lumbar spine suggests that the mechanism of injury is different between the two groups and therefore that DRI should not be used in both blast and ejection injury prediction.

	Ejection n (%)		Blast n (%)		P value	Relative Risk Blast/Ejection
Cervical	4	(2)	21	(27)	<0.001	12.7
Thoracic	93	(49)	42	(54)	0.5041	1.1
Lumbar	24	(13)	55	(70)	<0.001	5.6
C1	0	(0)	4	(5)	0.0069	N/A
C2	0	(0)	6	(8)	0.0005	N/A
C3	0	(0)	3	(4)	0.0243	N/A
C4	0	(0)	3	(4)	0.0243	N/A
C5	0	(0)	3	(4)	0.0243	N/A
C6	2	(1)	8	(10)	0.0011	9.7
C7	0	(0)	12	(15)	0.0001	N/A
T1	0	(0)	8	(10)	0.0001	N/A
T2	0	(0)	5	(6)	0.0019	N/A
T3	2	(1)	6	(8)	0.0088	7.3
T4	7	(4)	10	(13)	0.0106	3.5
T5	12	(6)	12	(15)	0.0316	2.4
T6	21	(11)	12	(15)	0.4132	1.4
T7	24	(13)	9	(12)	1	0.9
T8	28	(15)	9	(12)	0.5622	0.8
T9	27	(14)	11	(14)	1	1.0
T10	22	(12)	11	(14)	0.6828	1.2
T11	30	(16)	7	(9)	0.1735	0.6
T12	40	(21)	13	(17)	0.5002	0.8
L1	29	(15)	32	(41)	0.0001	2.7
L2	7	(4)	27	(35)	0.0001	9.3
L3	4	(2)	28	(36)	0.0001	17.0
L4	2	(1)	22	(28)	0.0001	26.7
L5	1	(1)	16	(21)	0.0001	38.8

Table 25: Ejection vs Blast Injury - number of victims with a fracture at each level or zone, relative risk and P value by Fisher's Exact Test between mounted blast victims in this thesis and ejection victims in the published literature. Relative risk = Risk in blast / Risk in ejection. There is a statistically significant difference at most levels [65, 114, 124, 132].

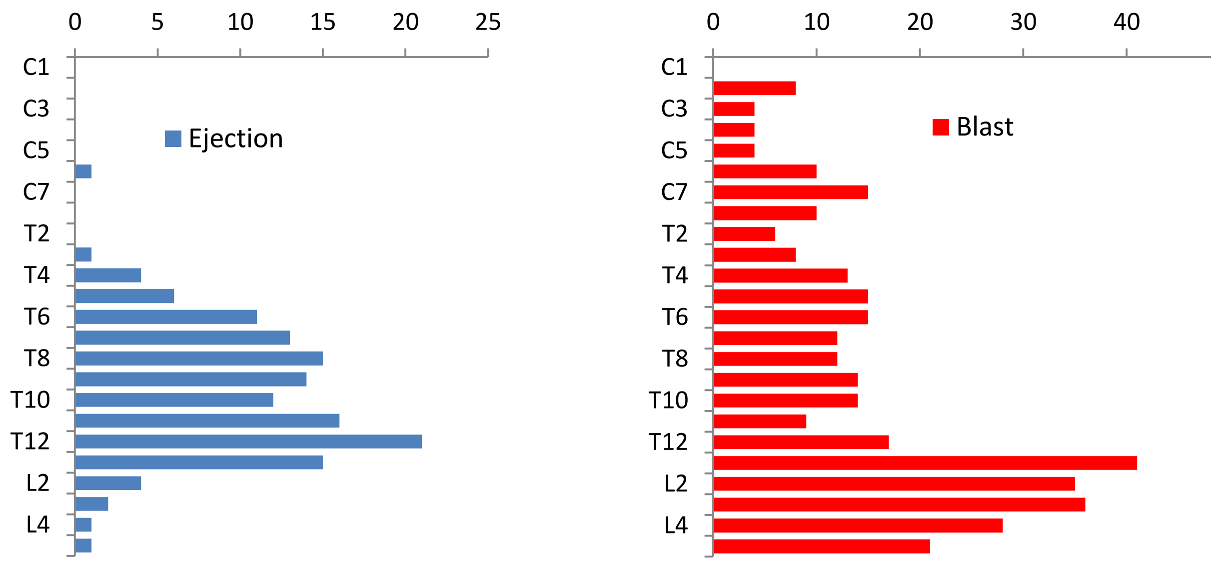

Figure 42: Percentage of injuries at each spinal level in ejection and blast in the study population. The different injury patterns are clear.

6.5 Summary

Since the test standard used for vehicle design in underbody blast protection is based on that used for aircraft ejection seats, this chapter reviewed the principles of ejection seats and sought to confirm whether a single injury prediction model is appropriate to both scenarios.

Aircraft ejection seats were developed to improve aircrew ability to escape from fast jet aircraft, and depend on a combination of explosive charge and rocket propulsion to drive a seated pilot upwards and out of the aircraft. This involves a significant axial load and a resulting risk of spinal injury.

Comparing the injury patterns in UK blast victims with those seen in aircraft ejection shows that the distribution of injury is markedly different. It is therefore unlikely that the mechanisms of injury are similar, so a single injury prediction model is not appropriate for both applications. This suggests that, even if DRI is adequate for aircraft ejection incidents, it cannot be applied equally to underbody blast. The seating posture of a blast victim is probably different to that of an ejecting pilot, the acceleration is significantly different and more variable, and the axis of load may be different. Each of these must be accounted for in a blast injury risk model, which must be based on a realistic understanding of the behaviour of the spinal column.

The next chapter will take an important first step in allowing a better model to be developed, by linking the mechanistic hypothesis of spinal fracture in blast elucidated from the UK blast cohort with an *in vitro* study of the thoracolumbar spine under high rate loading.

7. The effect of posture on fracture patterns in a simulated underbody blast loading scenario

7.1 Introduction

This thesis has introduced the injuries seen in underbody blast. It has discussed some of the methods used to reduce the risk of injury to vehicle occupants when a vehicle is struck by blast, and briefly introduced the tests used to evaluate them. In order to establish a useful model of the spine following underbody blast, it is important to understand how each part of the spine behaves when exposed to blast load. This thesis has shown that thoracolumbar injuries are common, and appear to be related to a flexion posture and movement. This chapter therefore seeks to establish a link between posture and fracture pattern when the spine experiences the rates of loading expected during blast.

The loading rates encountered in underbody blast are not clear. Manufacturers and governments do not publish details of seat acceleration in blast tests for security reasons. Additionally, the mass of the vehicle, size and location of the exploding device, soil type and charge burying depth, and seat position and type all have an unknown effect on the effect of a blast on each vehicle occupant. The occupant also may have an effect; the occupant's torso mass may alter the spine's response to axial load, and this may be further affected by body armour and personal equipment. There are therefore many uncontrolled and unknowable variables that affect the load applied to the spine by underbody blast. This chapter attempts to establish a reasonable range of values for the loads that might be seen by a victim of underbody blast.

7.1.1 Previous Studies

Yoganandan *et al*. established a technique for comparing loading rate with injury pattern [197]. In this paper, lumbar spine specimens comprising T12-L5 were mounted on a drop tower with a load cell and a torso mass. The specimens were dropped on to polyurethane energy absorbing material from one of three heights, with an impact velocity of up to 5.4 ms^{-1}. Peak axial forces of up to 3 kN were produced. Some thoracic specimens were also tested. In this study, there was no attempt to report the posture of the vertebrae at the moment of failure. The fractures produced were not clearly described, but included multifragmentary vertebral body fractures and lamina fractures. This study therefore associated impact velocity with failure load, but failed to consider the posture of the test specimen or the effect of load and posture on fracture pattern.

Langrana *et al.* [107] performed motion-segment studies in a materials tester using human cadaveric specimens, though all were over the age of 60. Thoracolumbar specimens were loaded in a neutral or extended posture at a rate of 0.1 ms^{-1}. All the specimens demonstrated crush fractures of the vertebral body. However, a low loading rate such as this in specimens likely to be affected by age-related osteoporosis may not be applicable to a military context with young patients and a higher loading rate. This study therefore does not inform the question posed in this thesis.

Stemper *et al.* [181] examined the rate-dependent fracture characteristics of L2-L4 lumbar vertebral bodies. Their experiments, similar to Yoganandan [197] – indeed Yoganandan is a co-author on this paper – dropped lumbar spine specimens with a 32 kg torso mass supported above. A load cell measured peak loads and high-speed video photography was used to measure displacement between vertebrae. Most of their specimens were loaded at under 2 ms^{-1} impact speed, though their highest rate was 4 ms^{-1}. Peak force through the specimen was found to correlate with impact speed. The specimens demonstrated a mix of wedge and burst fractures, but the authors did not relate the injury pattern to different loading rates or postures, so the critical challenges in this thesis remain unanswered.

The effect of posture on compressive strength under quasi-static loading was investigated by Adams [8]. In this study, a complex roller system was used to control the posture of a single lumbar spine motion segment. The study showed that increasing flexion up to 20° had a minimal effect on failure strength of the vertebra. The fracture patterns resulting from these tests were discussed briefly. This important paper established a link between posture and both fracture pattern and load to failure, but at too slow a loading rate for this thesis. Additionally, the complex system used to control posture would probably not produce reliable and repeatable results at higher loading rates.

These papers show that, although there is some evidence relating loading rate to failure load and posture to fracture pattern, there is no published work which controls both posture and loading rate and there is no work assessing the effect of posture on fracture patterns at a high loading rate. The aim of this chapter therefore is to seek to correct this deficiency, and to develop a simple, reliable and repeatable experiment that would be applicable at the loading rates seen in blast.

7.1.2 Hypothesis

In the reviewed literature and in the clinical data from this thesis, the most commonly injured segment of the spine in underbody blast is the thoracolumbar junction. Chapter 5 suggests that different segments of the spinal column experience different fracture patterns and postulates that this may relate to the posture of that section of the spine, with flexed spines experiencing more vertebral body wedge compression and burst fractures, and extended parts of the spine experiencing posterior element injuries.

This thesis suggests that, at higher loading rates such as those seen in blast, the fracture pattern in a thoracolumbar specimen will depend on the posture at the moment of impact.

This study therefore examines the effect of posture on injury patterns in the T12 vertebra. Specimens including T11-L1 were selected in order to allow simulation of different postures of the thoracolumbar junction and full mobility of the T12 vertebra of the specimen. Specific objectives of this study are to establish any link between posture and fracture pattern, and to see if there is a difference in the load to failure at different postures. This study also aims to develop a reliable, simple and repeatable method for testing thoracolumbar spine specimens at high loading rates.

7.2 Methods and materials

Ethical approval for this study was obtained from the local regional ethics committee (Imperial College Healthcare Tissue Bank and REC Wales; approval 12/WA/0196). The cadaveric specimens were provided by a licensed tissue laboratory (Life Legacy, Arizona, USA) and tissue donors had consented to their use for scientific research. All the experiments were conducted at Imperial, in a laboratory designed and approved for human tissue experiments in accordance with the Human Tissue Act 2004.

7.2.1 Specimen preparation

Specimens were received as fresh frozen, infection-screened, whole spines. They underwent computed tomography (CT) scanning (Siemens Somatom Definition AS 64, Erlangen, Germany) at the Centre for Defence Radiology, Royal Centre for Defence Medicine, Birmingham UK to exclude any pre-existing degenerate disease or structural abnormality. They were stored in a tissue freezer at -20°C and were thawed overnight prior to testing.

The required specimen was carefully dissected out of the remainder of the spine by the author. Ribs were removed by disarticulation. T11 and L1 discs were separated from the caudal and cranial endplates using sharp dissection and endplate remnants cleared with blunt dissection to maintain integrity of the vertebral body, disjointing T11 to L1 from the rest of the spine to leave a bisegment specimen. Facet joints were disarticulated at the end vertebrae but left intact at the central vertebra. The anterior longitudinal ligament (ALL), posterior longitudinal ligament (PLL), ligamentum flavum (LF), and interspinous and intertransverse ligaments were divided by sharp dissection and cleared from the ends of the specimen. Paraspinal muscles were removed, leaving all interspinous ligaments and periosteum intact. Small areas of periosteum were then cleared to allow strain gauge application.

Specimens were instrumented with strain gauges (C2A-06-062LW0350, Vishay, Basingstoke, UK) to identify potential location and time of fracture, the last taken to be immediately after peak strain was reached and strain began to reduce. They were aligned with the cranio-caudal axis of the specimen, perpendicular to the vertebral end plates, at two points on each side of the vertebral body and on either side of the spinous process (Figure 44), based on the technique used by Langrana [107]. Six gauges were used in each specimen. The bone was defatted with 100% ethanol solution at the site for gauge application, dried thoroughly, and gauges placed and secured with cyanoacrylate. They were then covered with silicone solution in order to prevent dehydration. The potted specimens were protected from drying during the experiment by regular water spray and wrapping in damp towels.

Strain gauges and the load cell were connected to a PXIe data acquisition system using customised LabVIEW software (NI Instruments, Austin, TX, USA). Data were collected at 25 kHz. High speed photography (Phantom V210 camera) was used to film the specimen during impact at 13,000 frames per second for an overview of the experiment to estimate the time of failure, examine the relative movement between vertebrae, and to ensure that no technical errors occurred.

7.2.2 Design of test rig

Previous studies in the spine [181, 197] have established techniques for supporting a vertebra in a metallic pot using bone cement, and this straightforward technique was used in this study. It was necessary to devise a means of controlling the posture of the motion segment in the pot. Based on the range of flexion and extension postures in Adams' study [8], this experiment aimed to test in 20 degrees of flexion and 10 degrees of extension for each specimen, which meant that each vertebral level would be 10° flexed or 5° extended. For these experiments, plastic wedges 5 mm in height were 3D printed with a 5° or 10° wedge angle (Figure 47). Straight blocks were also printed. During cementing, these were placed between the pot and the vertebral endplate to ensure that each vertebral body was potted at the correct angle. These blocks were designed by the author to provide a simpler way of controlling posture than the system of rollers used in Adam's paper.

If a specimen was potted at both ends using a wedge, it would be challenging to ensure that the two pots were aligned and parallel at the point of impact. During cementing the specimens were held parallel to each other using a steel plate screwed to the pot and held parallel by three M8 threaded rods (studs) secured with nuts and checked with an electronic level (Figure 43); this technique was proposed by a supervisor. To control the posture at impact, the author designed a plastic ring that was positioned on top of the corner of the top pot and secured using M6 studs to the drop tower base plate (Figure 45). The pots were therefore secured level and directly in vertical alignment, but the ring disengaged from the pot at the moment of impact so did not have an effect on the motion of the system.

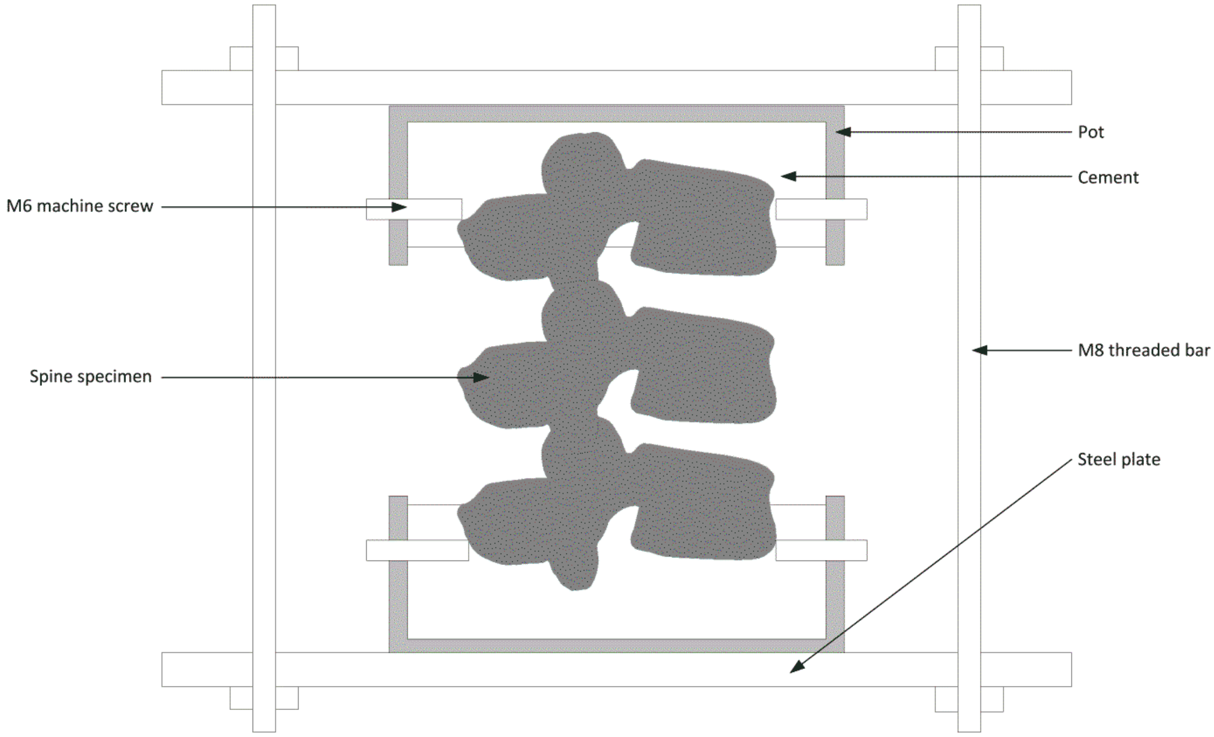

Figure 43: Controlling specimen alignment during cementing. The pots and specimens are held with aluminium plates at top and bottom. The two plates are held relative to one another with M8 threaded rods; before the second pot was filled with cement, the level was checked with a digital spirit level.

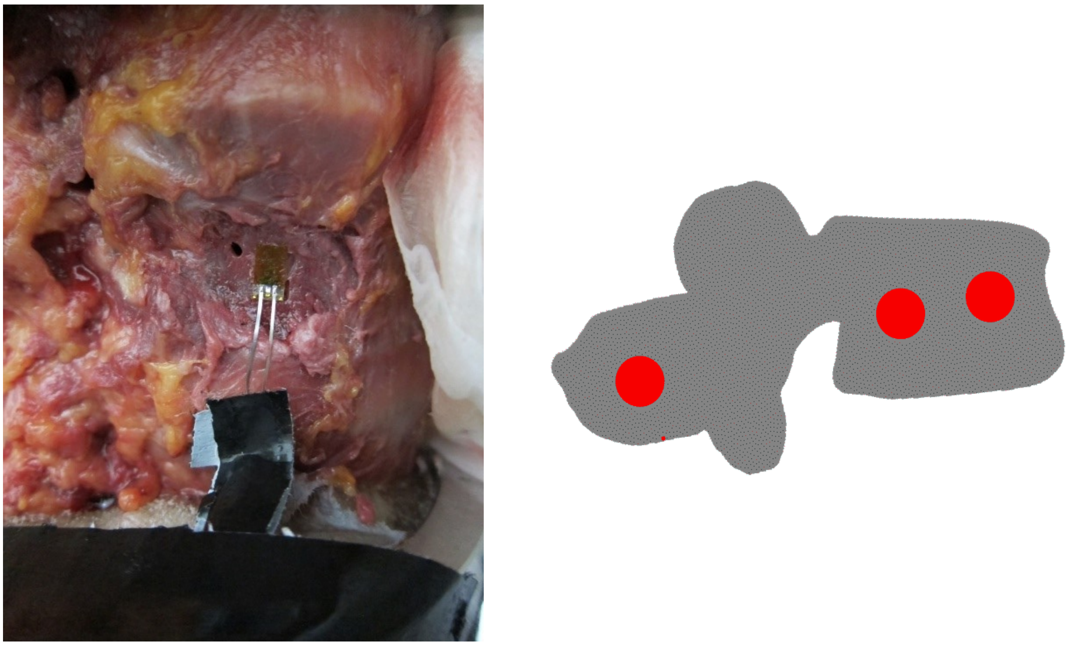

Figure 44: Location of strain gauges for drop rig tests. Left, gauge in situ on potted specimen. Right, side view of vertebra showing location of three gauges as red circles; the pattern was repeated symmetrically on the other side.

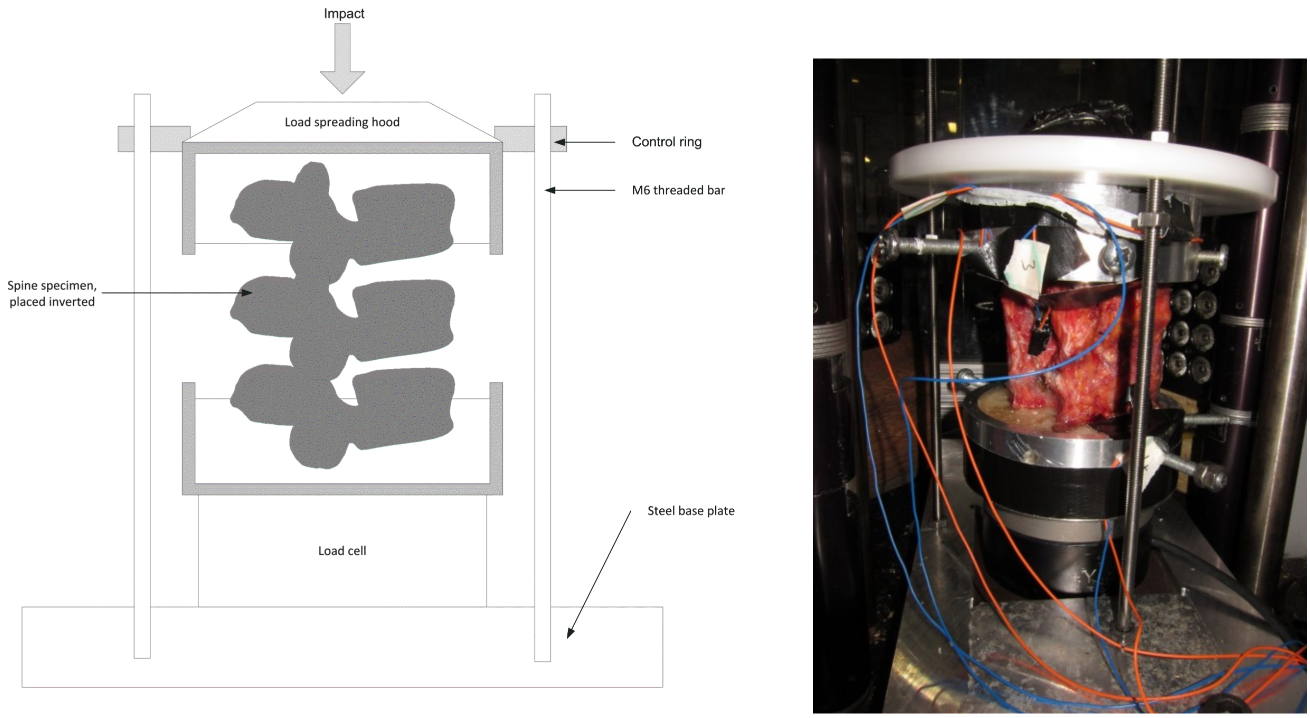

Figure 45: Drop tower arrangement showing posture control ring designed to ensure the specimen was axially aligned prior to tests. The plastic ring is secured with threaded rods to the plate on which the load cell is placed. Before impact, the ring was adjusted to level using the nuts on the threaded rod and a digital spirit level. Because the ring did not have a close fit with the side of the top pot, as soon as the rig impacted the pot, the ring disengaged from the system and had no effect on the experiment.

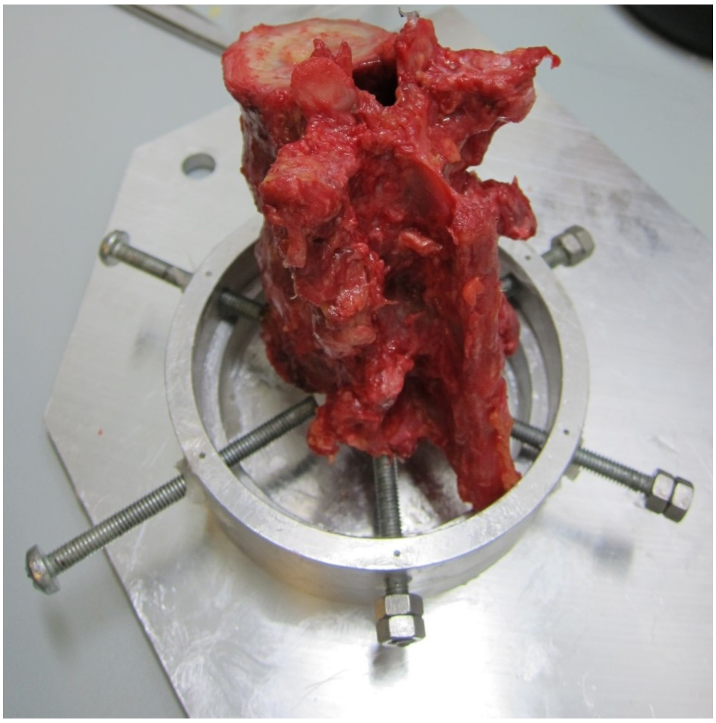

Figure 46: Stabilising two level spinal specimen for cementing. The specimen is placed centrally in the pot, resting on the position control wedge to ensure the correct alignment (not visible here), and held in place with M6 screws.

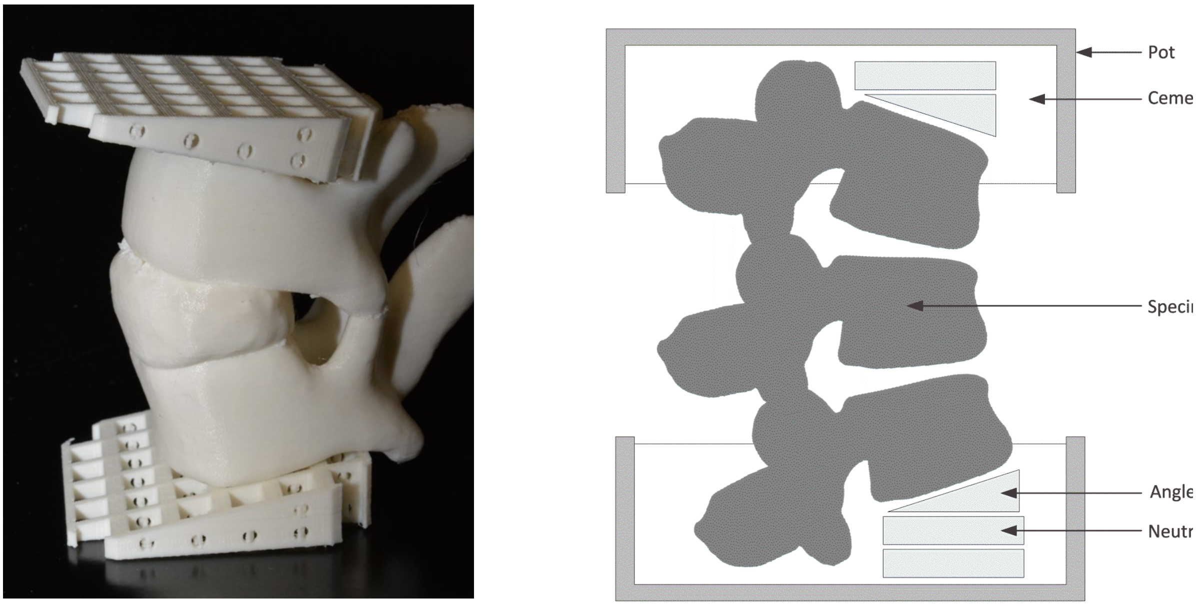

Figure 47: 3D printed posture control wedge in position with a printed trial motion segment, and schematic diagram demonstrating the use of the posture control wedges during cementing.

7.2.3 Testing Protocol

Specimens were placed in a cylindrical aluminium pot (Figure 43) with the endplate sitting on a plastic block or wedge to control the posture (Figure 47). For neutral position tests in 0° flexion, a flat block was used to set the height of the vertebral body within the pot; angled wedges were used for flexed and extended tests. M6 screws were used to stabilise the specimen for cementing. These screws were applied with the minimum possible force and did not disrupt the specimen's integrity. The specimen was held between two pots with plates secured with threaded bar, held parallel as shown in Figure 43. Each pot was then filled with polymethyl methacrylate (PMMA) bone cement so the vertebral body was almost completely covered. Once the cement had set, the specimen was inverted and the other end potted using the same protocol.

The specimens were placed in a drop tower (Instron Dynatup 9250, Instron, High Wycombe, UK) with a multiaxial load cell (Sunrise Instruments, USA) beneath the specimen. The specimens were placed inverted to ensure that the impulse from the drop tower acted in the same direction as underbody blast. Automatic pneumatic brakes were used, firing at the moment of impact, to prevent multiple contacts between drop rig and specimen.

Prior to testing with human tissue, the experimental apparatus, protocol, and data acquisition were tested using 3D printed surrogate specimens.

7.3 Results

7.3.1 Preliminary tests

To calibrate the equipment and confirm the impact speed, a T9-T10 motion segment was tested initially. The specimen was loaded as described and subjected to repeated drops at increasing rates until fracture. 45 minutes were allowed to pass between tests to allow soft tissues to relax. The drop heights, impact speeds, and results are at Table 26..

Drop height (m)	Speed at impact (ms^{-1})	Peak load at load cell beneath specimen (N)
0.04	0.88	420
0.08	1.25	1000
0.45	2.97	5580

Table 26: Results of preliminary tests using T9-10 specimen at three different loading rates. This test evaluated the measurement methodology and supported the selected impact speed.

The 0.04 m test was repeated with and without the posture control ring; no difference was noted in the peak load, confirming that the ring was not interfering with the rig following impact. The preliminary test specimen was found to have a superior burst fracture of T9 on scanning. No soft tissue injury was identified on dissection.

7.3.2 Impact testing in flexed, neutral, and extended posture

Four specimens were available for testing at different postures, as shown in Table 26. All were male and under the age of 60. Each was subject to a single impact at the same rate.

Test Number	Specimen Age	Posture	Drop height (m)	Impact velocity (ms^{-1})
1	53	Neutral	0.64	3.5
2	58	Extension	0.64	3.5
3	49	Flexion	0.64	3.5
4	43	Neutral	0.64	3.5

Table 27: Posture test data showing the age, position, drop height and impact velocity for the main tests. Measured impact velocity was consistent.

7.3.3 Post-test examination

Following testing, each specimen was scanned using the same protocol as the pre-impact scans to identify the resulting injury. Scans were assessed by both the author and a consultant radiologist with conventional radiology software. Specimens were then dissected to confirm the radiological findings and inspect the discs and ligaments for any soft tissue injuries that may have been missed by the scan.

7.3.4 Fracture patterns

All four specimens demonstrated burst fractures (Table 28). There were no spinous process fractures. The first neutral specimen demonstrated a superior incomplete burst fracture with a bilaminar fracture, representing a two-column injury resulting from axial compression according to the Magerl classification (Figure 48 and Figure 49) [118]. A similar pattern was seen in the second neutral specimen (Figure 54~~Figure 54~~ and Figure 55). The extended posture specimen also showed a superior incomplete burst fracture but with no posterior element involvement (Figure 50 and Figure 51). Again, the burst fracture was entirely in the posterior part of the vertebra. The flexed specimen also showed a superior incomplete burst fracture (Figure 52 and Figure 53). Here, however, the anterior vertebral body was also involved, suggesting that force here was greater than in the neutral specimen; this is confirmed by the experimental measurements presented below.

Specimen	Scan findings	Additional findings from dissection
1 (Neutral)	Posterior superior incomplete burst with bilaminar fracture, <5mm retropulsion, ligaments intact, no kyphosis	Soft tissues intact
2 (Extension)	Posterior superior incomplete burst, no lamina or spinous process fracture, ligaments intact	Facet joints more mobile at T11-T12, but not disrupted
3 (Flexion)	Superior burst fracture of T12, no retropulsion, no posterior involvement, no ligament injury	Soft tissues intact
4 (Neutral)	Posterior superior complete burst with right laminar fracture and no retropulsion.	Soft tissues intact

Table 28: Imaging and dissection findings for the three definitive drop test specimens. There is a consistent finding of a vertebral body fracture with slightly different pattern in each posture.

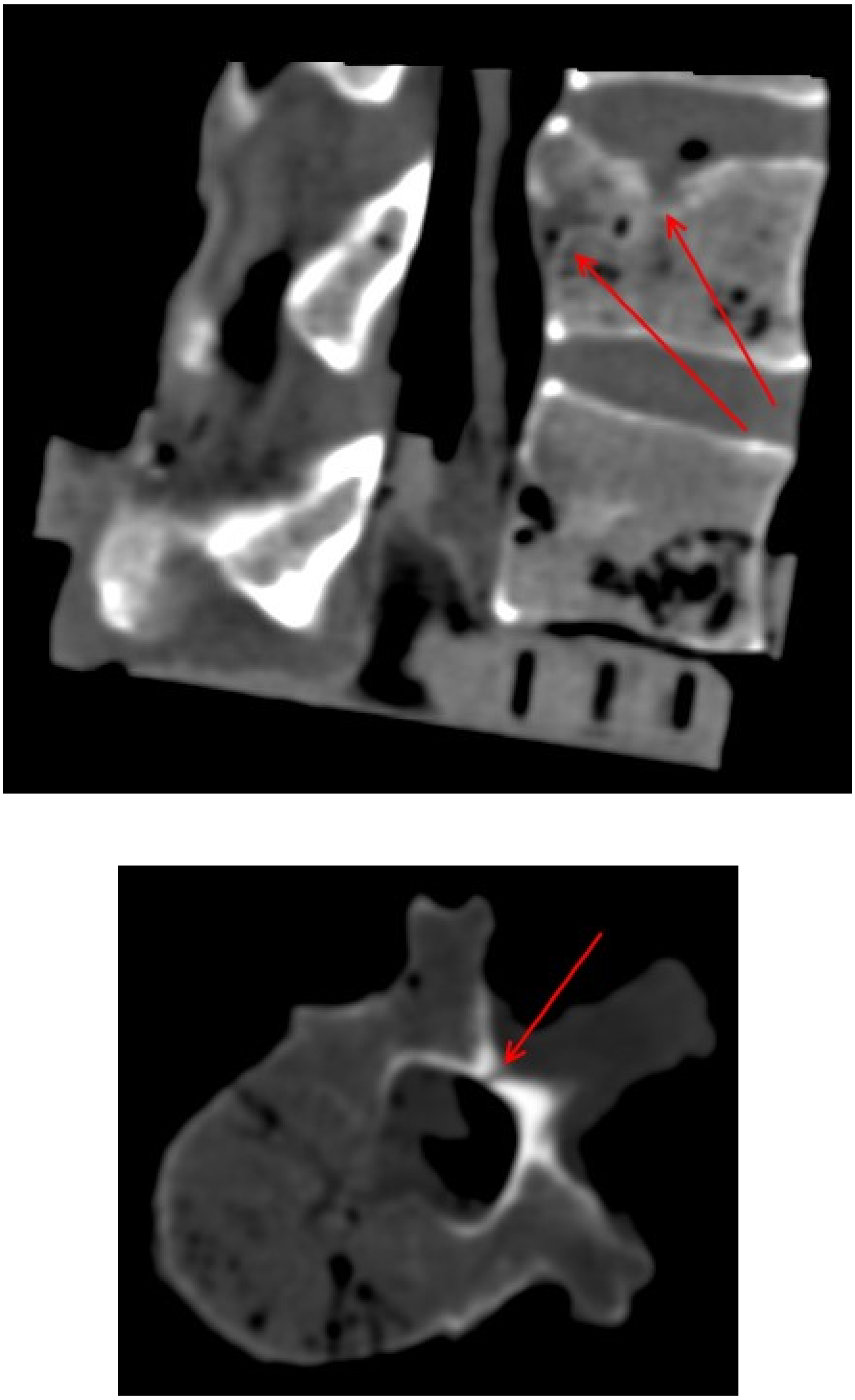

Figure 48: Post-test specimen 1 (neutral posture), sagittal view CT scan, showing burst fragment in T12 highlighted by red arrow. The second image shows the lamina fracture. Note that the body fracture is exclusively in the posterior body. The apparent breach in the body cortex on this image is a blood vessel.

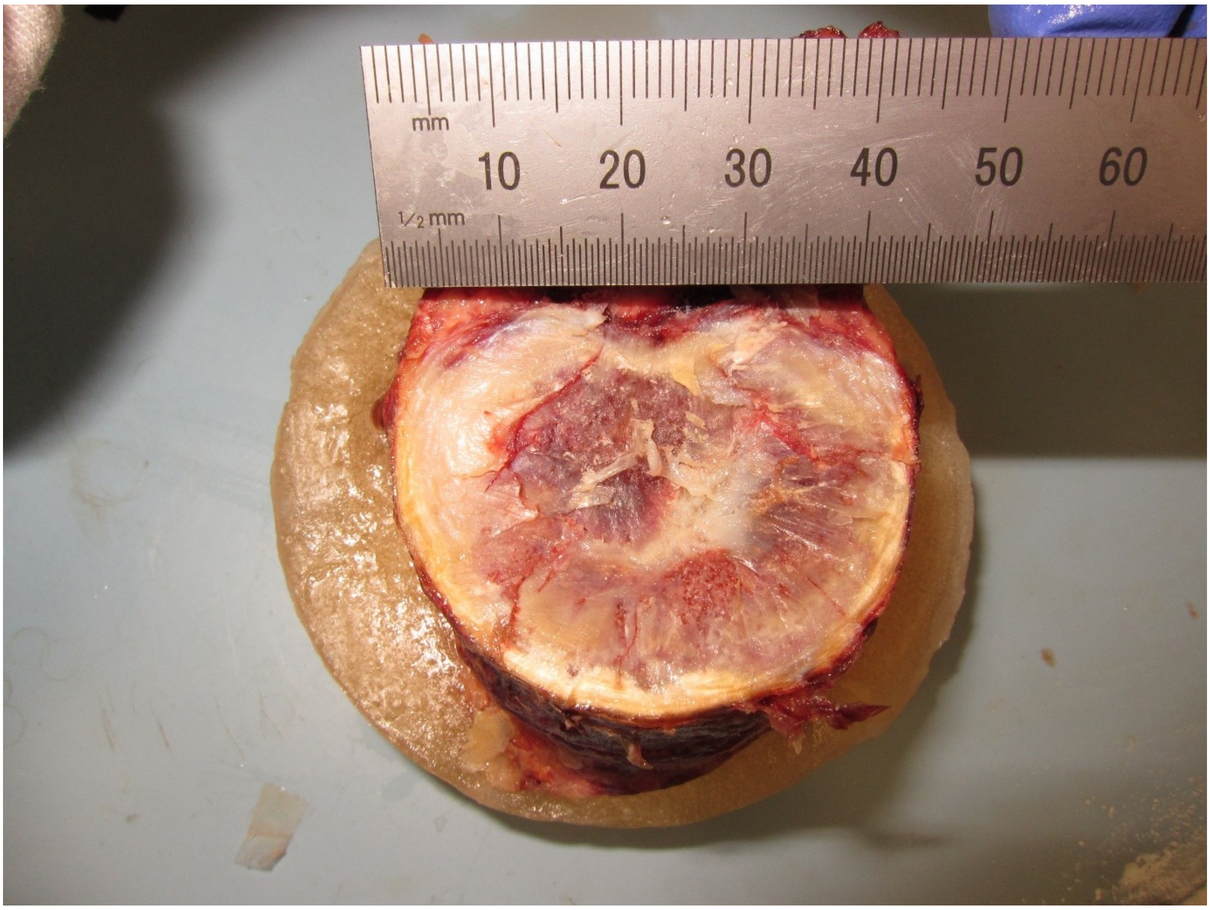

Figure 49: Specimen 1 post-test dissection image. The superior vertebral body is shown, with fracture lines visible.

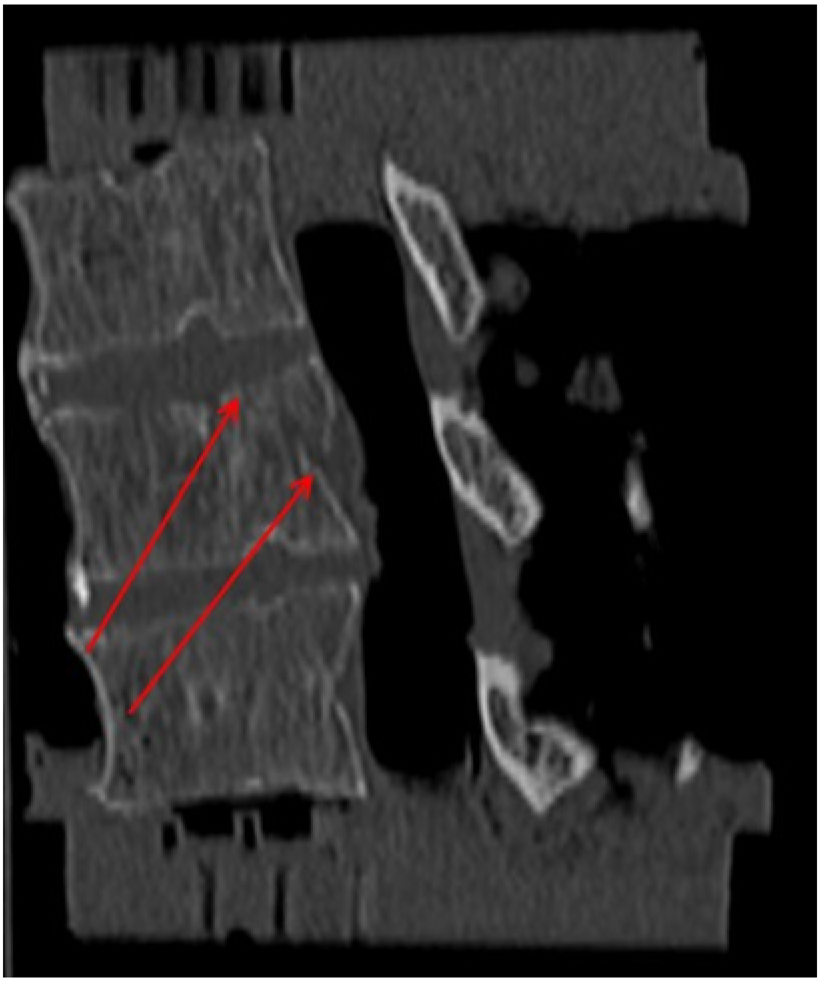

Figure 50: Post-test scan of Specimen 2, tested in an extended posture, showing the burst fragment in T12 at the posterior part of the vertebral body.

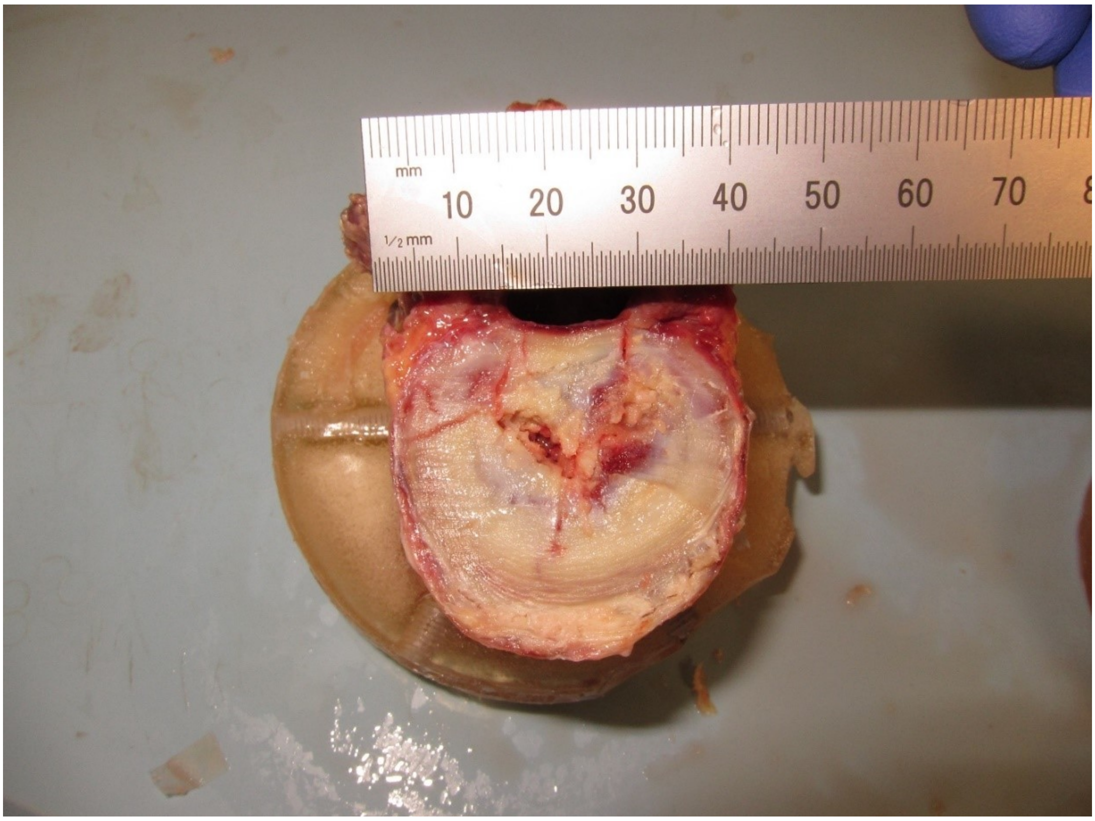

Figure 51: Specimen 2 following post-test dissection. Note the fracture lines extending to the posterior vertebral body.

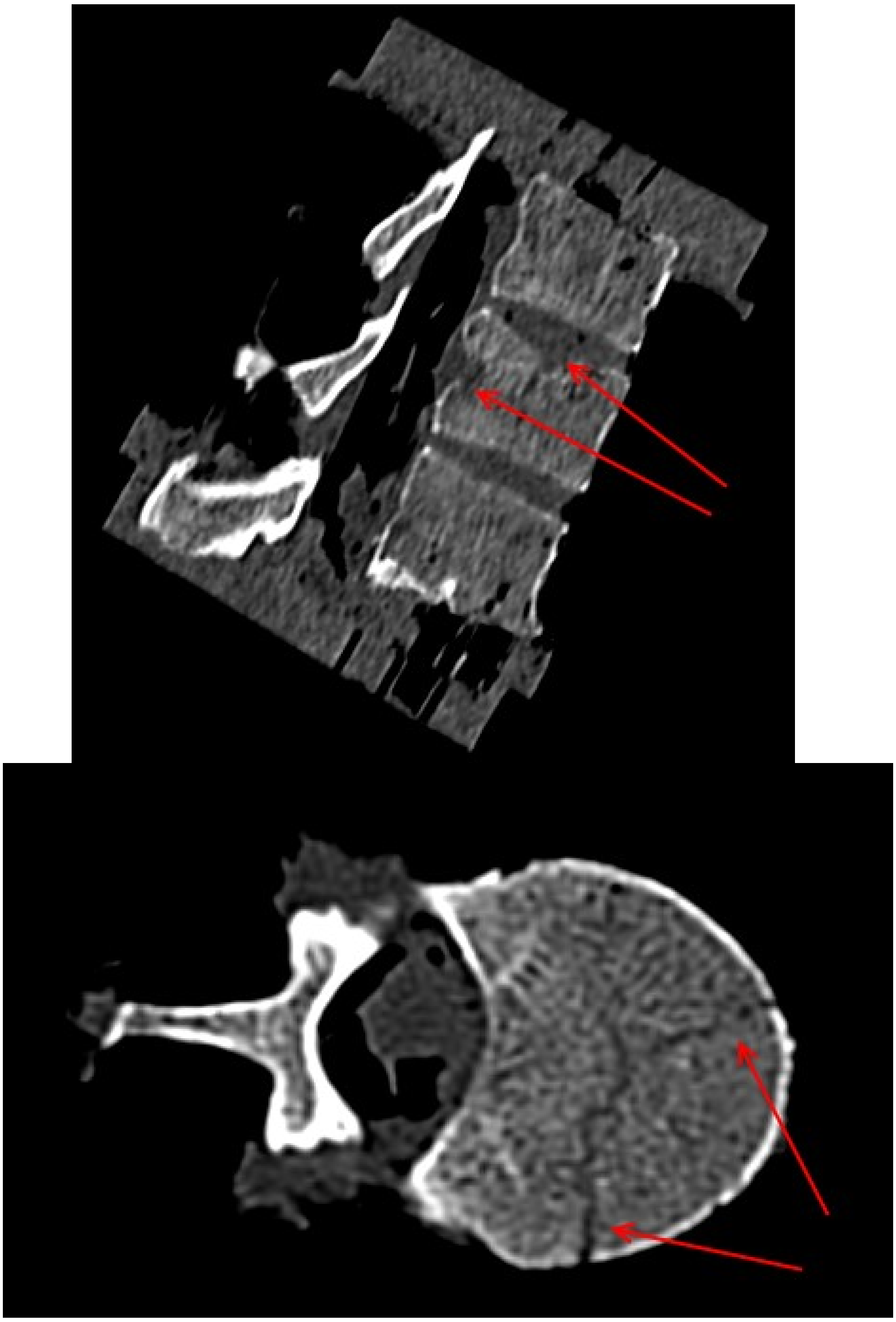

Figure 52: Post-test scan of Specimen 3, tested in flexion, showing the burst fracture extending to front of vertebral body

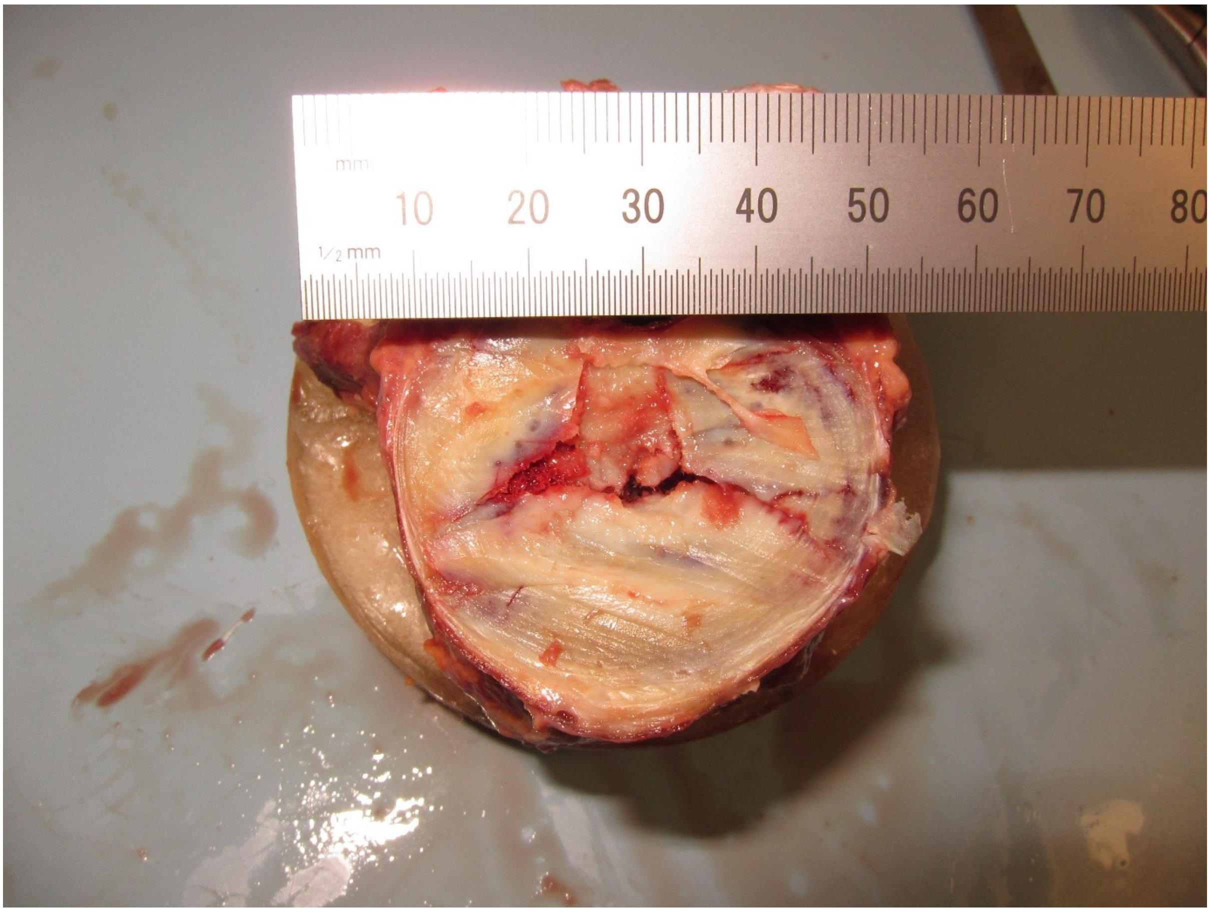

Figure 53: Specimen 3 post-test dissection, showing fracture lines extending to the front of the vertebral body

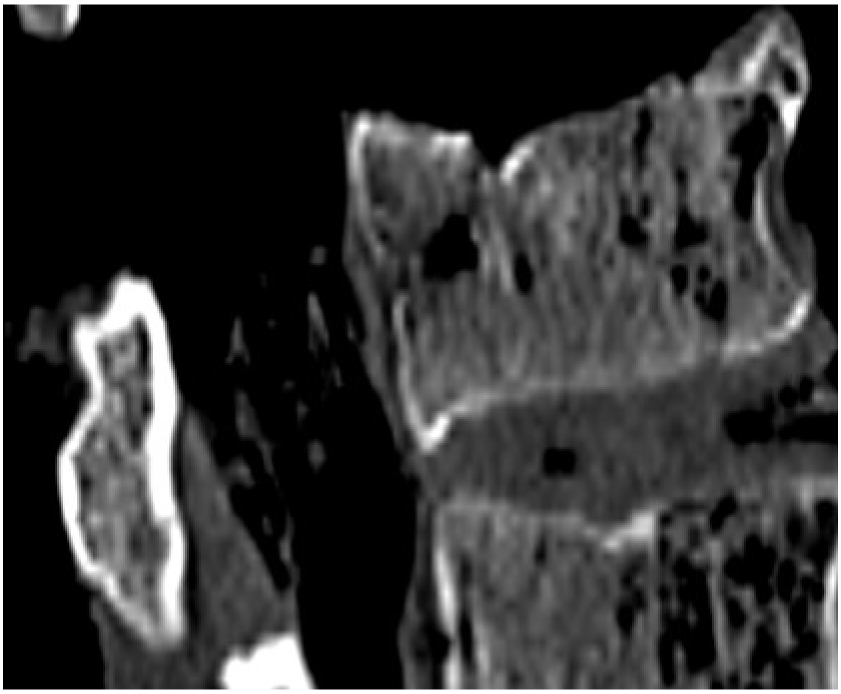

Figure 54: Specimen 4 post-test scan, showing a superior posterior fracture following a neutral test

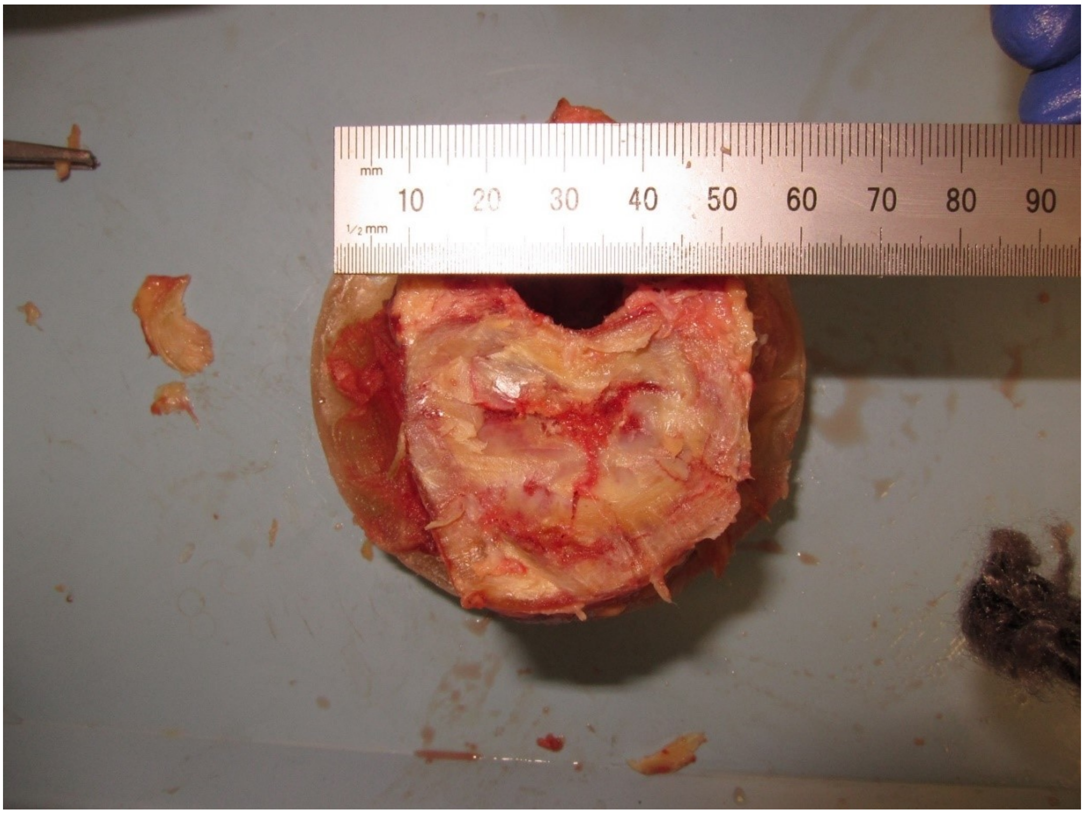

Figure 55: Specimen 4 post-test dissection, with a similar fracture pattern to Specimen 1

7.3.5 Video Analysis

In this study, high speed video was primarily used to ensure that no unintended kinematics / loading were occurring, such as impact between the fixing screws and posture control ring. No such errors occurred. The system was also used to identify the movement of the specimen as it approached failure.

In both of the neutral test specimens, the vertebral body fractures in axial compression and the motion segment then moves in to extension. In the flexed specimen, the body fails while flexing, and the motion segment rotates in to extension after failure as the specimen “flattens out” The extended specimen exhibited minimal movement at the posterior elements and posterior vertebral body, compressing in its extended posture.

7.3.6 Load to failure

Table 28 shows the peak load at failure and strain shown at different postures.

Specimen	Peak load at failure (N)	Body strain at failure (%)	Spinous process strain (%)
1 (Neutral)	9292	-1.3	-0.24
2 (Extension)	13478	-0.75	-0.11
3 (Flexion)	9753	-0.3	Gauges failed
4 (Neutral)	8209	-0.7	-0.04

Table 29: Impact test data at different postures showing the differing load to failure and strain in a flexed specimen.

The key finding is that the specimen in an extended posture failed at a significantly higher load than the flexed or neutral specimens. The load against time curve for each specimen is shown in Figure 56.

7.3.7 Strain

As shown in Table 28, the strain gauges on the spinous process were not reliable. Figure 56 shows the peak strain at vertebral body failure, demonstrating the time of vertebral failure. Further interpretation was not possible due to the unreliability of the measurements.

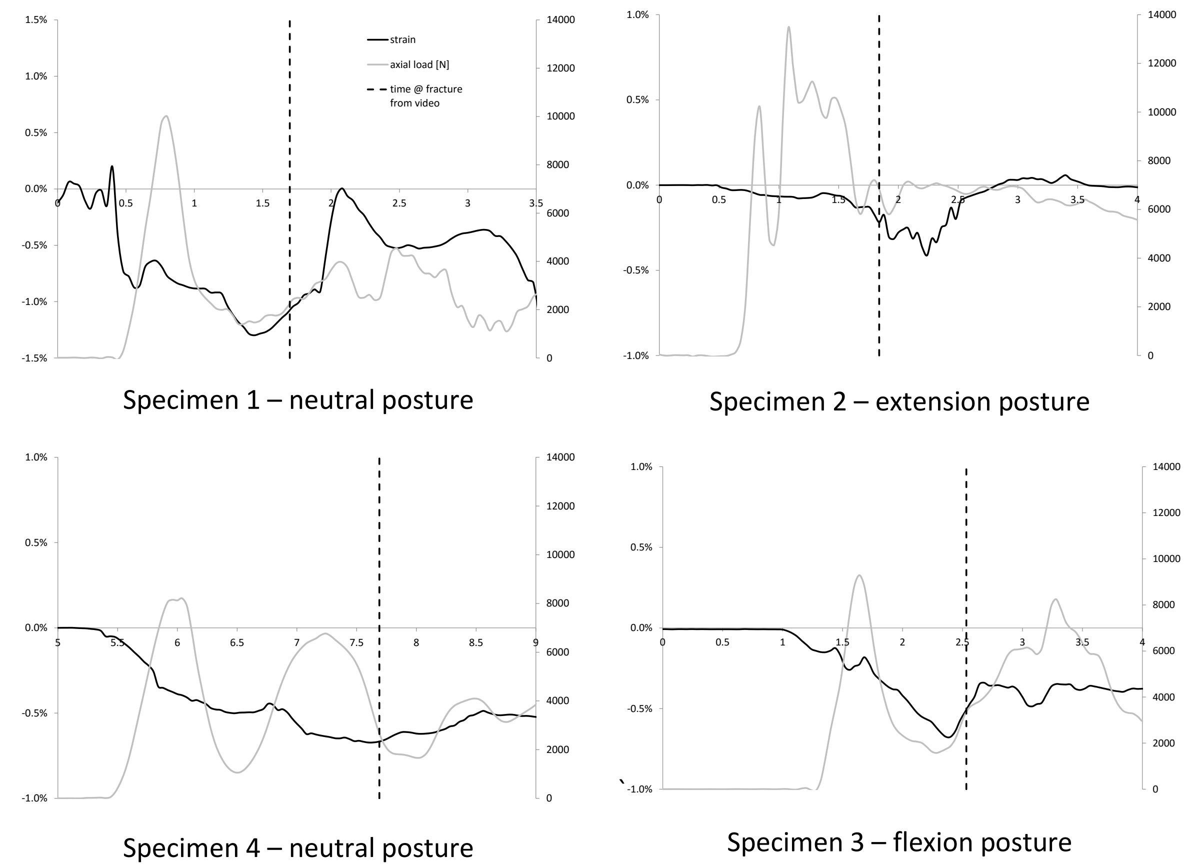

Figure 56: Axial load (right axis, grey), vertebral strain (left axis, black), and time at fracture as estimated from the high speed video recording (dotted line) for the 4 specimens. Each graph shows a peak load reached just before the moment of fracture.

7.4 Discussion

7.4.1 Preliminary work

In developing the technique for this experiment, some preliminary experiments were carried out and a few challenges overcome. As this thesis describes, previous papers have used complex systems to control alignment when assessing the effect of spinal column posture on fracture pattern. Previous authors have used cement potted specimens to successfully test high loading rate fracture behaviour. A simple method of controlling posture was therefore sought. The author suggested a system of 3D printed plastic wedges, manufactured at the selected posture and placed between the test specimen and a pot. It was felt that this technique would ensure the correct posture in each test as long as the wedge was securely placed on the vertebral endplate. In order to trial this technique, as well as the proposed load cell arrangement and error-free data acquisition, a selection of spine models was 3D printed from CT data by research colleagues. In preliminary tests the wedges were successful and no problems encountered.

The second challenge was the need to control the axial alignment of the whole specimen up to the moment of impact. The author suggested a plastic posture control ring which could be used to ensure that the top specimen was level and the specimen axially aligned up to the moment of impact. This method was tested in preliminary thoracic spine tests, with repeated tests at different rates to confirm that the device did not affect the load seen by the system throughout testing.

7.4.2 Selection of Loading Rate

The acceleration seen at the vehicle seat, and therefore at the lumbar spine, following blast is not clearly known. Alem *et al.* tested an energy absorbing seat, recording 32g acceleration in 25 ms after a blast test with a peak velocity of 7 ms^{-1}. The vehicle floor is suggested to accelerate rapidly to over 12 ms^{-1}, so this velocity has been used in lower limb experiments [120, 153]; however it is unclear whether the same acceleration is seen at the seat pan, or at the spine as a result of the acceleration of the lower limbs and of the deforming and vehicle floor and vehicle movement.

Papers detailing experiments described as high loading rates include compressive loading between 1.2 ms^{-1} in a hydraulic system [54] and 5.4 ms^{-1} with a 32 kg mass drop test [197]. In the last paper, all specimens failed when impacted at 5.4 ms^{-1} but each had been tested at two lower rates, so microdamage may have accumulated; this may not have been picked up by radiographs between tests. Stemper *et al*. [181] in a similar paper noted fractures occurring with a single impact at under 4 ms^{-1}. A target loading rate of 3.5 ms^{-1} was therefore selected for this experiment. In this experiment a 7 kg mass was selected to provide peak loading of over 5 kN, shown in our preliminary test to lead to fractures.

7.4.3 Selection of Posture

Although the overall range of motion of the lumbar spine is well known, the individual range of motion of each vertebral motion segment is not well documented. Adams [8] described the range of motion of specimens with a wide range of flexion from 6 to 20.9° (mean 12.4°). Since their experiments confirmed a small effect on compressive strength at 9° flexion, a flexion posture of 10° was selected for our experiments. The mean extension range in their series was 4.5°, so 5° was selected for this study as it was felt that a smaller angle would be difficult to control.

This is the first attempt to study vertebral fracture patterns and mechanics at a known loading rate while controlling posture. As discussed, the loading rate encountered at an individual spinal level during a blast event is unknown; further, different vehicle characteristics, charge sizes and detonation locations mean that there is likely to be a wide range of loading rates seen in the spine during blast and one test cannot reproduce them all. Nonetheless, it was felt that this loading rate was sensible and within the range seen in reality.

7.4.4 Comparison with Previous Findings

It was intended during this study to repeat each experiment. Unfortunately, the tissue supplier was not able to provide enough suitable specimens within the time available, so only one test was performed in each condition. Nonetheless, some useful data emerge, with the fracture pattern moving from posterior to anterior as the specimen moves from extension to flexion. This is consistent with previous findings by Hoshikawa *et al.*, who used quasistatic tests to show that the posterior spine is loaded in progressive tension as flexion angle increases [85].

Compared to Yoganandan and Stemper's work, there are some slight differences in fracture pattern in this study [181, 197]. Stemper's study showed a more typical anterior wedge or burst fracture pattern than the posterior burst shown in these tests. It may be that this is because in this study the vertebral body was not precisely centred under the impact point, but the impact point was closer to the spinal canal. However, Yoganandan's study showed similar fracture patterns to these tests.

In this study, despite the small sample size, there is a difference in the fracture pattern with changing posture. A flexed posture may lead to more anterior involvement in the vertebral body than a neutral or extended posture. This suggests that there may be a relationship between posture at the moment of injury and fracture pattern, and that this pattern is similar to that described in the civilian literature and classifications. This gives some confidence that the mechanistic conclusions drawn from the clinical cases in this thesis are likely to be accurate.

In these experiments, the peak load at failure was higher in extended posture than in neutral. Adam's paper [8] which tested the load to failure in spinal specimens at different postures did not find a difference in load to failure at different postures. It is perhaps possible that the change in load-sharing between disc, body, and facet joints as the vertebra moves from flexion to extension is responsible for this change, but repeated measurements are needed in order to draw this conclusion. The effect the difference in peak load to failure has on fracture risk for a given impact at different postures is unclear, as the pattern of failure is so different. DRI depends on an assumption that fracture risk is proportional to load, and if this is not the case it might help explain why DRI is unreliable.

Examining the high speed video suggests that in these test the initial load path is through the vertebral body and disc. As the flexed or neutral vertebral body moves in to more extension, the axial load recorded by the load cell falls, perhaps because the posterior facet joints share more load as the motion segment rotates in to extension. In the extended posture, this did not happen. Repeated tests are necessary to confirm this suggestion.

The effect of loading rate on fracture pattern at these high rates is unclear. It is possible that different fractures would be produced by repeating these experiments at higher and lower rates; this is work for the future. This small study has shown a difference in fracture pattern when the posture is altered. This contributes to the understanding of how the spine fails during an underbody blast event, as it supports the mechanistic hypothesis drawn in previous chapters: as the spine flexes, the fracture pattern involves more vertebral body injury and a more tensile failure of the posterior spine; and as the spine moves during blast, the thoracolumbar junction flexes, producing similar fracture patterns in clinical studies as in this *in vitro* study. Therefore, modifying the seating posture in vehicles may help control the risk of injury. Further repeats of these experiments are needed in order to confirm this proposition.

8. Clinical Outcomes in Spinal Injury due to Blast

8.1 Introduction

As this thesis has shown, blast victims often have spinal injuries. However, they tend to be part of a pattern of multiple trauma. Although there are many published papers describing blast injuries, they do not focus on the spine; those papers which detail spinal injury describe injury patterns, mechanisms and epidemiology, but not clinical outcomes. Only three published series attempt to describe clinical outcome in spinal injury due to blast and these lack detail.

Rivera et al. [156] published an epidemiological paper describing long term disability in soldiers discharged from the military following combat injury with a focus on the spine. Vertebral body fractures were the most common injury in the series, and most were caused by blast. Details of the injuries and time to discharge are lacking, but the study suggests that blast related spinal fractures are an important problem leading to morbidity and disability.

Freedman et al. [63] reported a case series of thoracolumbar burst fractures following underbody blast which is appropriate to the subject of this thesis. Sixty-five patients were reported with a mean age of 29.7. Only two were female. 80% had vertebral fractures between L1 and L3. Eight patients had more than one spinal fracture. 54.8% of victims had a lower limb or pelvic fracture. 43% had some degree of neurological deficit. 68% had surgery for their fracture, and 18% of those experienced surgical complications such as infection. Of the patients with a neurological deficit who underwent surgery, 11% improved. However, long term outcomes and function were not reported.

Cozen [47] reported a series of 15 people who had back pain following exposure to blast in World War Two. No diagnosis was recorded and no imaging carried out. A history of back pain was associated with persistent symptoms with follow-up of eight days. The paper mentions that all patients had lumbar pain, consistent with the incidence of lumbar fractures in later series and this thesis.

Schneider et al. [162] reported long term outcomes in spinal cord injured soldiers from World War Two. The study reports the presence of long term pain in those with a closed spine injury, but no details of the injuries or incidence of symptoms are given.

Recent individual case reports have highlighted the devastating nature of spinal blast injuries, such as multiple open lumbar fractures with lumbosacral dissociation [95] and an L5 burst fracture with good function following fusion [96]. These reports, however, are short papers with minimal detail and lack functional outcome scores.

The complexity of spinal injuries in blast was highlighted by Possley and reviewed by Bernstock [22, 146]. They highlighted that spinal fractures rarely occur in isolation, which confounds the ability to identify spine related clinical outcome. However, these papers did not describe spinal injuries in detail, although they did show that blast victims with spinal injuries also tended to have other injuries, many of which led to complications such as infections; so the injury burden in each victim is complex and the effect of their individual injuries on their function may be hard to unravel.

The published literature, therefore, do not give a clear impression or the effect spinal fractures from blast have on their victims in terms of pain and function. This is partly because blast victims often have complex multiple injuries other than their spinal injuries. There is therefore a clear need for such a study as identifying the clinical and functional effects of these injuries will identify which injuries are most disabling, thus helping guide mitigation strategies to prevent the most important injuries and help direct treatment strategies.

8.2 Hypothesis

Victims of blast injury with multiple injuries may have particular clinical effects, and therefore allow identification of key injuries that may later be amenable to mitigation strategies.

8.3 Aim

This chapter aims to identify the clinical outcomes and functional outcomes in victims of blast-related spinal fractures.

8.4 Method

This study received ethical approval from the Ministry of Defence Research Ethics Committee (protocol 519/MODREC/14). All victims identified in the injury pattern analysis in this thesis were considered for this study, thus identifying a patient cohort of 71 survivors. There are a few more victims in the injury pattern analysis, who were identified after this outcome study had been performed. These survivors were screened by clinicians involved in their care to exclude those with significant post-traumatic stress disorder, in whom a questionnaire study was felt to be likely to worsen their symptoms, and head injuries such that they were unlikely to be able to complete a questionnaire. Where a current address could be identified for the remainder, a questionnaire study was sent by the author.

The questionnaire, shown in Annexe 3, included details of return to duty, return to driving and sport, and hobbies. The Spinal Cord Independence Measure [35] was used to assess for neurological functional deficit; the Oswestry Disability Index as originally described by Fairbank [60, 61] was used to assess the level of pain, and the Short Form 36 survey was used to assess the effect of injuries on daily living [188]. The modified Fritz Oswestry index was not used. Sexual function was assessed with a single question. These questionnaires were selected because they are used in comparable military and civilian papers, so would provide a straightforward way to compare this study with other data. The SF36 questionnaire was used to gain an overall picture of function and health, which might be helpful in trying to tease out the spinal injury from the victim's overall injury burden.

Injury and demographic data were taken from the database established in Chapter 5. ISS, NISS, and associated injuries were identified. Clinical records were reviewed to identify surgical procedures.

Data were analysed using SPSS software (IBM). Given the small numbers of respondents, outcomes were converted to ordinal "normal" or "abnormal" scores and a logistic regression model was used to identify indicators of a poor score. P values less than 0.05 were taken to be significant.

8.5 Results

8.5.1 Overview

Of the 71 survivors in the series, 6 questionnaires were returned with consent (Figure 57). The mean age was 27.5 (range 20 – 35) and mean time post injury was 40 months (21-75). All were male. Three were mounted.

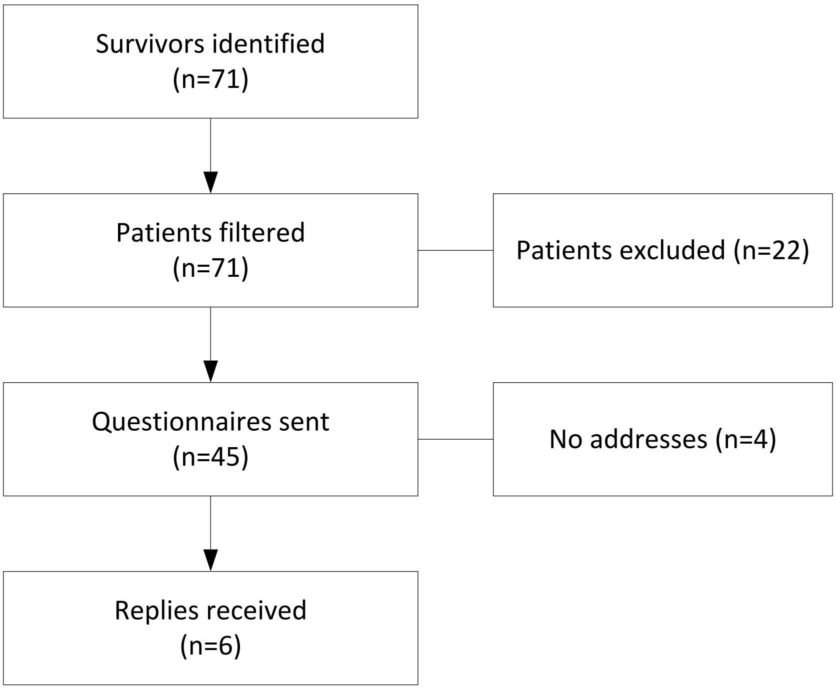

Figure 57: Patient questionnaire, consent and reply flow chart showing the number of questionnaires sent out and the replies received.

There were five cervical, two thoracic and nine lumbar fractures. One victim had a cervical fusion procedure but there were no other spinal interventions. One victim had bilateral transtibial amputations, one had a calcaneal and humeral fracture, and one had a tibial fracture and calcaneal fracture treated surgically.

Mean time for return to duty was 18 months (8-26) with 2 not back at work. Mean time for return to driving was 5 months (2-12). Only two had returned to hobbies including fishing, running, and motorcycle racing; the others were all unable to run. All three victims with associated injuries felt that the non-spinal injury delayed their return to driving, duty, sports, and hobbies more than the spinal injury.

8.5.2 Spinal Cord Independence Measure

The Spinal Cord Independence Measure (SCIM) [35] reports the results of separate questions relating to function. In each case, a score of 0 implies the greatest degree of pain or dependence and the highest score is normal function in that category.

All patients reported normal function in self-care and transfers. One required a stick for walking and stairs, and a leg orthosis. One required assistance with wheelchair transfers. One required touch for sexual stimulation, but all others reported normal function. All had normal bladder and bowel function. It was not possible to calculate significant predictors of a poor score in the SCIM due to the number of identical values. Summary outcome measures are reported in Table 30.

8.5.3 Oswestry Score

The Oswestry score ranks function and pain questions from 0 to 5, with 0 being pain free and independent function [50, 60, 61]. The total score expressed as a percentage implies the degree of disability – up to 20% suggests "minimal disability", up to 40% suggests "moderate disability", and up to 60% is described as "severe". Two patients reported a score over 20%.

8.5.4 Short Form 36 Outcome

The SF-36 questionnaire is different in that it references patient responses to a population norm of 50, so scores under 50 are "below normal" and scores above 50 are "better than normal" [188]. Compared to the population norm, the SF-36 scores were within 2% of normal in all domains. The physical component summary score was 48% and vitality score 48%, with normal patients scoring 50%. As the SF-36 scores were narrowly distributed and close to a normal score in all cases, no significant predictors of a poor outcome could be identified.

Score	Mean	Min	Max
SCIM Self Care (20)	19.7	18	20
SCIM Respiration (18)	18	18	18
SCIM Mobility (40)	36.8	32	38
Oswestry Index	12.7	0	32
SF36 Physical Function	49.2	38.4	57.5
SF36 Role Physical	48.9	39.2	57.2
SF36 Body Pain	51.5	42.6	62.0
SF36 General Health	47.7	29.4	61.7
SF36 Vitality	47.6	37.7	58.5
SF36 Social Functioning	49.8	27.3	57.3
SF36 Role Emotional	54.4	45.7	56.2
SF36 Mental Health	51.3	35.2	61.3

Table 30: Summarised Outcome Measures. Maximum scores in brackets. For the SF36 scores, a "normal" function score is 50. There is significant variability in responses.

8.6 Discussion

This study has attempted to evaluate the functional and clinical outcomes in blast related spinal injury. Unfortunately, very few victims responded to the study, so only limited conclusions can be drawn in such a small number. It could be argued that with so few respondents the statistical analysis is invalid. However, the author felt that the respondents were owed a debt of gratitude for participation and that the best possible analysis should be attempted on the available data; it would have been disingenuous to the participants to have left the data out of this thesis.

It is only possible to speculate reasons why the response rate was so poor, but it is assumed that some addresses were no longer correct and it is known that this population is frequently asked to participate in studies so it may be the case that the victims are suffering from "questionnaire fatigue". Further, the low response rate may be associated with some bias, from the lack of responses from those too ill to answer, and from responses by those who are particularly happy or unhappy with their outcomes.

Should this study be repeated, other strategies might be used to improve the response rate including face to face or telephone meetings, or focus groups where victims could collectively draw on their experience. However, ethical approval for this study did not provide these options.

The poor response rate means that this study does not allow conclusions to be drawn. In these respondents, most reported good functional outcomes and limited pain. There were no specific features, other than age, which predicted a sub-optimal score in the analyses used, but it is recognised that these are of limited utility in samples of this size.

There are no comparable military series published. There are several larger civilian studies of spinal trauma outcomes, but given the limited data from this study it is difficult to compare them. Additionally, not all civilian studies use the same outcome measures as this series. Ko [105] reported a series of thoracolumbar burst fractures treated with posterior instrumentation and reported similar Oswestry scores to this paper, although all Ko's patients had isolated spinal injury. A similar trend is seen in the series of burst fractures reported by Defino [53], where the SF-36 outcomes, while not reported in detail, are similar to those in this series.

Although this is a very small series, it shows that the victims of blast have multiple injuries including spinal fractures and that the spinal injuries contribute a degree of pain and functional limitation in most cases. A larger study would be needed to establish trends and predictors of poor outcome.

9. Summary, Impact and Future Work

9.1 Summary

The Improvised Explosive Device, along with military anti-vehicle mines, represents a significant hazard to military and civilian personnel alike, whether involved in peacekeeping, mine-clearance or humanitarian operations. IED attacks on vehicles lead to devastating injuries, including spinal injuries, but there has been little research on the mechanism and outcome of spinal injuries in blast. This thesis therefore established a need for better understanding of the mechanism of injury to the spine from underbody blast, and the effects of such injury on their victims.

When an IED detonates beneath a vehicle, a high pressure shock wave accelerates towards the vehicle, carrying a mass of soil and fragments with it. The shockwave and fragments strike the vehicle, causing the vehicle to deform and accelerate. Force is transmitted to vehicle occupants both by the accelerating vehicle and the deforming floor; this combination of mechanisms is known as "solid blast".

This thesis reviewed the literature surrounding the patterns and mechanisms of injury in underbody blast. The published literature provides little detail, but suggests that spinal injuries are caused by a mixture of axial load along the line of the spine and movement of the lumbar and cervical spine.

Chapter 5 reviewed injury patterns in both mounted and dismounted UK IED victims. It is clear from this review that there is a consistent pattern of injury in mounted blast victims, combining axial force along the spine with flexion at the thoracolumbar and thoracic-cervical junctions as well as flexion at the thoracic apex. This is the first comprehensive review aiming to explain the mechanism of spinal injury in blast; it has furthered understanding of the behaviour of the whole spinal column during underbody blast. In the cervical spine, it suggests a buckling mode of failure, as the central cervical vertebrae consistently fail anteriorly, in a flexion pattern, along with an association between cervical spine injuries and skull fracture, which implies that the cervical spine buckles between the skull and the immobile thorax. The injury patterns in the thoracic spine are now more clearly understood, with association between thoracic vertebral body fracture and adjacent spinous process fractures, combined with the patterns of vertebral body fracture, supporting the notion that the thoracic spine fails in compression and flexion. In the lumbar spine, thoracolumbar fractures occur in flexion and lower lumbar fractures are a result of axial load, perhaps because they are closer to the axis of a load applied through the vehicle seat. These suggest that thoracic and lumbar fractures are caused by a predominantly axial load through the vehicle seat, but as the lower limbs are thrown upwards by the deforming vehicle floor the fracture pattern is altered.

Since armoured vehicles started to develop, designers have sought to improve their resistance to underbody blast and reduce the injury burden to occupants when such attacks occur. Aircraft ejection seats are also associated with a risk of spinal injury and seat designers tried to understand the behaviour of the spine during axial acceleration in order to reduce this risk. Chapter 6 reviewed the injury prediction models in current use and demonstrated that they are unsatisfactory for the complex scenario of underbody blast, because the injury patterns and mechanisms in blast and ejection are significantly disparate. While the mitigation strategies in current use may be effective, they could be improved with a better understanding of the interaction between vehicle, seat, and occupant; this depends on improving understanding of the behaviour of the spine in blast in order to create a better injury prediction model.

Having established a mechanistic hypothesis for the behaviour of the thoracolumbar spine during a blast event, Chapter 7 developed an experimental model of the spine to test the effect of posture on fracture pattern. As the spine moves from flexion to extension, the involvement of the anterior vertebral body in fractures caused by high-rate loading reduces. This supports the loading hypothesis suggested in Chapter 5, namely that the moving lower limbs flex the lumbar spine and control the fracture pattern at the thoracolumbar junction.

This understanding of the effect of posture may not only support design of a new injury prediction model, but may in the short term aid vehicle designers in altering the posture of a seated vehicle occupant to moderate the fracture pattern if an underbody blast attack occurs. In the sections of the spine subject to a load anterior to the spine while the spine was flexed, significant flexion-compression injuries occurred. Where the spine was loaded axially, burst fractures were the most common pattern. Clinically, a wedge-compression injury is likely to have less significant sequelae than a burst fracture. Possibly, by altering the seated posture to reduce the exposure of the spine to axial and bending loads - for example by tilting the seat backrest back a few degrees - these injuries could be made less likely. Mitigation strategies to reduce the blast load transmitted to the lower limbs will also have a positive effect on the fracture pattern.

The clinical effect and long-term outcome of blast related spinal fractures has not been reported in the literature. Chapter 8 attempted to correct this with a questionnaire based outcome study of UK victims of underbody blast. However, the response rate was poor and the sample size very small. In this study, most blast victims had other severe injuries such as amputations, but did not have the most severe spinal injuries identified in the UK blast victim population. This small study demonstrated that where victims of blast have both spinal and limb injuries, their spinal injuries may not be the most significant cause of disability. However, if victims of more significant spinal injury had been included, this may have altered the outcome. A further study is being carried out by the Defence Medical Service looking at the long term morbidity outcomes in all blast victims – the ADVANCE study. This may help resolve this important question, because identifying the most important blast injuries in the spine will allow those injuries to be targeted by mitigation strategies in vehicle design.

9.2 Future Work

9.2.1 Understanding the spine in blast

This clinical series has improved understanding of the mechanism of injury in underbody blast. The experimental study has supported understanding of some of the most common injuries. However, we still do not understand what the effect of loading rate on fracture pattern is, and an understanding of the behaviour of the whole spine is required.

The first question to address is the effect of loading rate on injury pattern. Initially, the experimental study performed for this thesis must be continued to include repeated tests at different loading rates. It may also be possible to repeat the tests at intermediate flexion and extension postures. This will elucidate the link between loading rate, posture, and fracture pattern. It will also be useful to feed the data in to an evolving finite element model of the spinal column for use in simulated blast tests in order to allow realistic simulated blast tests in future as well as supporting development of a useful underbody blast test dummy

Whole spine tests are therefore necessary to examine the buckling behaviour of the spine during blast in order to understand how the whole spinal column responds. For these tests, it is suggested that the AnUBIS [120] is adapted to test the entire spinal column. Tests will apply axial load through the pelvis and measure load and acceleration at key vertebral bodies in order to quantify load transmission, motion, location of fracture and the interaction of each part of the spinal column as a system.

The human spine is surrounded by muscles, which apply a resting tension to the vertebrae and attached ligaments, and which contract to protect the spine as load is applied. While it is thought that these do not play a part in a blast event as the time to peak load occurs usually faster than the time it takes for muscles to fire, their behaviour is not well understood and further research is needed. An accurate whole-spine test would demand a method of simulating the effect of the paraspinal and abdominal muscles. The abdominal and thoracic contents also play a part in controlling the spine's behaviour, as the recruitment of mass in dynamic events affects to force experienced by the tissues; this would need to be understood and accounted for.

Later, a lower limb flexion model may be included. The challenge in these tests will be simulating the effect of torso mass, abdominal muscles and the effect of viscera and the posterior spinal muscles, and pressure. In these tests, it will be possible to account for the effect of body armour, helmets, and different torso mass. Combining the clinical and experimental data in this thesis with proposed future work will support development of a computer model of the whole spine to simulate the effect of underbody blast, and therefore development of mitigation strategies.

9.2.2 Mitigation strategies

Work in the lower limb has already led to the adoption of new vehicle features, such as installing seats suspended from the roof rather than attached to the floor. This thesis raises a few possibilities; it seems that strategies to reduce lower limb injury might also limit spinal movement, and changing an upright to reclined seat may reduce the risk of severe injury. However, before mitigation strategies can really be improved, the work commenced in this thesis must progress and understanding of the spine in blast be better.

It has been suggested that personal equipment such as body armour, and ergonomic features such as seat harnesses, alter the behaviour of the spine in underbody blast by moving the transition from rigid to flexible spine, and by providing a fulcrum about which the spine can rotate. Perhaps modified body armour and better integration between vehicles and personal equipment could improve this.

9.2.3 Towards a new injury prediction model

There are several injury prediction models in existence, each based on a limited understanding of the behaviour of the spine in underbody blast. Current prediction of spinal injury risk in underbody blast is therefore unacceptably poor. The DRI may be satisfactory for aircraft ejector seats, but does not account for the changing posture of the spine as the victim moves after underbody blast or the natural frequency response of the spine at the loading rates seen in underbody blast. Even if the DRI is accurate, it cannot differentiate a minor and a devastating injury.

An ideal model for blast tests would be simple and based on easily measurable force or acceleration. However, a reliable simple model needs to be based on a good understanding of the response of the spine following blast load. The current models are simple, but are based on inaccurate assumptions about the spine's behaviour.

It may be possible, once a sophisticated predictive model has been developed, to simplify it in to a conceptual, lumped parameter model along the lines of the DRI. It may even be possible to retain the DRI equation but apply new coefficients to improve its accuracy. However, it is important to understand the way the spine behaves before simplifying it to a conceptual model if the model is to be satisfactory and valid. Later, a computer simulation of the whole spine for axial load, accounting for the rest of the body and personal equipment, may be feasible and potentially allow vehicle designers to select a less devastating injury as the most likely outcome of blast.

9.2.4 A future clinical outcome study

This thesis attempted to establish a clinical outcome study in underbody blast, but the response rate was poor. It is not clear how this can be improved, but it would be valuable to attempt to gain better knowledge of the functional effect on victims of spinal injuries in blast. It may be that combining a spinal study with larger studies would reduce the burden on respondents and increase the response rate. The ADVANCE study, already under way, should go, hopefully, some way to remedying this.

9.3 Conclusion

Spinal injuries following underbody blast are common, devastating, and poorly understood. Vehicle designers have incorporated a variety of features to try to reduce the risk and burden of spinal injuries in underbody blast, but there is no validated way of testing these features, and therefore no proof that these strategies are as effective as the could potentially be.

This thesis is the first attempt to combine clinical and experimental data to establish a model of the behaviour, and has gone some way to establishing a better understanding of the behaviour of the spine in underbody blast. A consistent injury pattern has been identified which supports a mechanistic hypothesis for the behaviour of the spine in underbody blast. Experiments have shown how this mechanism applies to individual vertebrae.

The hope is that the ground work laid by this thesis will, in future, be integrated in a better injury prediction model, helping improve vehicle design to reduce the injury burden from blast and save lives.

10. Annexe 1: Search terms used for literature review in Chapter 5

Adult

Back Pain/epidemiology*

Accidents

Accidents, Aviation/statistics & numerical data

Accidents, Traffic/statistics & numerical data

Adult

Afghan Campaign 2001-

Afghanistan

Aged

Aged, 80 and over

Biomechanical Phenomena

Blast Injuries/complications*

Blast Injuries/epidemiology

Blast Injuries/etiology*

Blast Injuries/physiopathology*

Blast Injuries/prevention & control

Blast Injuries/therapy

Blast Injuries/epidemiology

Bombs*

Cadaver

Cervical Vertebrae/injuries*

Cervical Vertebrae/physiopathology*

Female

Great Britain

Humans

Immobilization/instrumentation

Incidence

Iraq War, 2003-2011

Leg Bones/injuries

Lumbar Vertebrae/injuries*

Male

Manikins

Military Personnel/statistics & numerical data*

Models, Anatomic

Multiple Trauma/etiology

Neck Injuries/epidemiology

Neck Injuries/physiopathology*

Neck Injuries/prevention & control

Posture/physiology

Registries

Retrospective Studies

Spinal Fractures/epidemiology

Spinal Fractures/etiology*

Spinal Fractures/pathology

Spinal Fractures/physiopathology

Spinal Fractures/prevention & control

Spinal Fractures/therapy

Spinal Cord Injuries/epidemiology*

Spinal Injuries/epidemiology

Spinal Injuries/physiopathology*

Spinal Injuries/prevention & control

Stress Disorders, Post-Traumatic/epidemiology

Thoracic Vertebrae/injuries*

United States/epidemiology

Warfare

Weight-Bearing/physiology

Wounds, Gunshot/epidemiology

Young Adult

-

11. Annexe 2: Papers describing distribution of blast injury in warfare

Author	Year	Paper	Type of paper	Number of patients in series	Number of patients with spinal fracture
Barr.	1946	Solid blast personnel injury: a clinical study	Case series	50	9
Bell	2009	Military traumatic brain and spinal cord injury	Review of registry data	408	40
Belmont	2013	The nature and incidence of musculoskeletal combat wounds in Iraq and Afghanistan	Review of registry data	797	715
Bevevino	2013	Incidence and morbidity of concomitant spinal fractures in combat-related amputees	Review of registry data	226	29
Bilgic	2009	Burst fracture of the lumbar vertebra due to a landmine injury	Case series	1	1
Blair	2012	Military penetrating spine injuries compared with blunt	Review of registry data	598	598
Blair	2012	Spinal column injuries among Americans in the global war on terrorism	Case series	598	598
Comstock	2011	Spinal injuries after improvised explosive device incidents: Implications for tactical combat casualty care	Case series	372	29
Davis	2003	Distribution and care of shipboard blast injuries	Case series	52	6
Dougherty	2009	Battlefield extremity injuries in Operation Iraqi Freedom	Review of registry data	935	9
Eardley	2012	Spinal fractures in current military deployments	Review of registry data	57	57

Eskridge	2012	Injuries from combat explosions in Iraq	Review of registry data	4263	"Low incidence"
Freedman	2014	The combat burst fracture study	Case series	65	65
Helgeson	2011	Retrospective review of lumbosacral dissociations	Case series	23	23
Jankovic	1998	Spine and spinal cord injuries during the war in Croatia	Case series	96	80
Lehman	2012	Low lumbar burst fractures: a unique fracture mechanism sustained in our current overseas conflicts	Case series	32	39
Parsons	1993	Spine injuries in combat troops – Panama, 1989	Case series	252	6
Possley	2012	The effect of vehicle protection on spine injuries in military conflict	Review of registry data	549	549
Ragel	2009	Fractures of the thoracolumbar spine sustained by soldiers in vehicles attacked by improvised explosive devices	Review of registry data	12	16
Schoenfeld	2013	Spinal injuries in united states military personnel deployed to Iraq and Afghanistan	Review of registry data	7877	797
Schoenfeld	2013	Characterization of spinal injuries sustained by American service members killed in Iraq and Afghanistan: a study of 2,089 instances of spine trauma	Review of registry data	5424	2089
Schoenfeld	2012	Characterisation of combat-related spinal injuries sustained by a US army brigade combat team during Operation Iraqi Freedom	Review of registry data	29	8
Schoenfeld	2012	Epidemiology of cervical spine fractures in the US military	Case series	4048	4048

Turegano-Fuentes	2008	Injury patterns from major urban terrorist bombings in trains	Case series	512	25

Table 31: Key papers in this review[15, 16, 18, 23, 24, 26, 27, 44, 51, 56, 57, 63, 81, 90, 108, 139, 145, 149, 164, 166, 167, 169, 170, 185]

12. Annexe 3: Functional outcomes following spinal column blast injury: Questionnaires

Thank you for completing this questionnaire and helping us gather useful data on the effects of blast injury.

Study Number: «Study_Number»

Date Questionnaire Completed:

1. Have you returned to full duties following your injury?

Yes No

On what date?

2. Did another injury delay your return to work more than your spinal injury?

Yes No

If yes, please give details.

3. Have you returned to driving?

Yes No

On what date?

If not, did your spinal injury alone prevent you from driving?

4. Have you returned to all of your hobbies, sports and leisure activities?

Yes No

On what date?

5. What sports and activities are you able to do at the level you want?

6. What sports are you not able to do at the level you were before your injury?

Spinal Cord Independence Measure

How well can you perform the following tasks? These questions have been adapted from the Spinal Cord Independence Measure and the American Spinal Injuries Association Autonomic Standards Assessment Form.

Self-Care

1. Feeding (cutting, opening containers, pouring, bringing food to mouth, holding cup with fluid)

- Needs parenteral, gastrostomy, or fully assisted oral feeding
- Needs partial assistance for eating and/or drinking, or for wearing adaptive devices
- Eats independently; needs adaptive devices or assistance only for cutting food and/or pouring and/or opening containers
- Eats and drinks independently; does not require assistance or adaptive devices

2. Bathing (soaping, washing, drying body and head, manipulating water tap).

A. Upper body

- Requires total assistance
- Requires partial assistance
- Washes independently with adaptive devices or in a specific setting (e.g., bars, chair)
- Washes independently; does not require adaptive devices or specific setting (not customary for healthy people)

B. Lower body

- Requires total assistance
- Requires partial assistance
- Washes independently with adaptive devices or in a specific setting

- Washes independently; does not require adaptive devices or specific setting

3. Dressing (clothes, shoes, permanent orthoses: dressing, wearing, undressing).

A. Upper body

- Requires total assistance
- Requires partial assistance with clothes without buttons, zippers or laces
- Independent with clothes without buttons, zippers or laces; requires adaptive devices and/or specific settings
- Independent with clothes without buttons, zippers or laces; needs assistance only for buttons, zips and laces
- Dresses any clothes independently; does not require adaptive devices or specific setting

B. Lower body

- Requires total assistance
- Requires partial assistance with clothes without buttons, zips or laces
- Independent with clothes without buttons, zippers or laces; requires adaptive devices and/or specific settings
- Independent with clothes without buttons, zippers or laces
- Dresses (any clothes) independently; does not require adaptive devices or specific setting

4. Grooming (washing hands and face, brushing teeth, combing hair, shaving, applying makeup)

- Requires total assistance
- Requires partial assistance
- Grooms independently with adaptive devices
- Grooms independently without adaptive devices

Respiration and Sphincter Management

5. Respiration

- Requires tracheal tube (TT) and permanent or intermittent assisted ventilation (IAV)
- Breathes independently with TT; requires oxygen, much assistance in coughing or TT management
- Breathes independently with TT; requires little assistance in coughing or TT management
- Breathes independently without TT; requires oxygen, much assistance in coughing, a mask (e.g., peep) or IAV (bipap)
- Breathes independently without TT; requires little assistance or stimulation for coughing
- Breathes independently without assistance or device (**Normal breathing)**

6. Sphincter Management – Bladder

- Indwelling catheter
- Residual urine volume (RUV) > 100cc; no regular catheterization or assisted intermittent catheterization
- RUV < 100cc or intermittent self-catheterization; needs assistance for applying drainage instrument
- Intermittent self-catheterization; uses external drainage instrument; does not need assistance for applying
- Intermittent self-catheterization; continent between catheterizations; does not use external drainage instrument
- RUV <100cc; needs only external urine drainage; no assistance is required for drainage
- RUV <100cc; continent; does not use external drainage instrument **(Normal)**

7. Sphincter Management - Bowel

- Irregular timing or very low frequency (less than once in 3 days) of bowel movements
- Regular timing, but requires assistance (e.g., for applying suppository); rare accidents (less than twice a month)

- Regular bowel movements, without assistance; rare accidents (less than twice a month)
- Regular bowel movements, without assistance; no accidents

8. Use of Toilet (perineal hygiene, adjustment of clothes before/after, use of napkins or diapers).

- Requires total assistance
- Requires partial assistance; does not clean self
- Requires partial assistance; cleans self independently
- Uses toilet independently in all tasks but needs adaptive devices or special setting (e.g., bars)
- Uses toilet independently; does not require adaptive devices or special setting)

Mobility (room and toilet)

9. Mobility in Bed and Action to Prevent Pressure Sores

- Needs assistance in all activities: turning upper body in bed, turning lower body in bed,
- sitting up in bed, doing push-ups in wheelchair, with or without adaptive devices, but not with electric aids
- Performs one of the activities without assistance
- 4. Performs two or three of the activities without assistance
- 6. Performs all the bed mobility and pressure release activities independently

10. Transfers: bed-wheelchair (locking wheelchair, lifting footrests, removing and adjusting arm rests, transferring, lifting feet).

- Requires total assistance
- Needs partial assistance and/or supervision, and/or adaptive devices (e.g., sliding board)
- Independent (or does not require wheelchair)

11. Transfers: wheelchair-toilet-tub (if uses toilet wheelchair: transfers to and from; if uses regular wheelchair: locking wheelchair, lifting footrests, removing and adjusting armrests, transferring, lifting feet)

- Requires total assistance
- Needs partial assistance and/or supervision, and/or adaptive devices (e.g., grab-bars)
- Independent (or does not require wheelchair)

Mobility (indoors and outdoors, on even surface)

12. Mobility Indoors

- Requires total assistance
- Needs electric wheelchair or partial assistance to operate manual wheelchair
- Moves independently in manual wheelchair
- Requires supervision while walking (with or without devices)
- Walks with a walking frame or crutches (swing)
- Walks with crutches or two canes (reciprocal walking)

- Walks with one cane
- Needs leg orthosis only
- Walks without walking aids

13. Mobility for Moderate Distances (10-100 metres)

- Requires total assistance
- Needs electric wheelchair or partial assistance to operate manual wheelchair
- Moves independently in manual wheelchair
- Requires supervision while walking (with or without devices)
- Walks with a walking frame or crutches (swing)
- Walks with crutches or two canes (reciprocal walking)
- Walks with one cane
- Needs leg orthosis only
- Walks without walking aids

14. Mobility Outdoors (more than 100 metres)

- o Requires total assistance
- o Needs electric wheelchair or partial assistance to operate manual wheelchair
- o Moves independently in manual wheelchair
- o Requires supervision while walking (with or without devices)
- o Walks with a walking frame or crutches (swing)
- o Walks with crutches or two canes (reciprocal waking)
- o Walks with one cane
- o Needs leg orthosis only
- o Walks without walking aids

15. Stair Management

- o Unable to ascend or descend stairs
- o Ascends and descends at least 3 steps with support or supervision of another person
- o Ascends and descends at least 3 steps with support of handrail and/or crutch or cane
- o Ascends and descends at least 3 steps without any support or supervision

16. Transfers: wheelchair-car (approaching car, locking wheelchair, removing arm and footrests, transferring to and from car, bringing wheelchair into and out of car)

- o Requires total assistance
- o Needs partial assistance and/or supervision and/or adaptive devices
- o Transfers independent; does not require adaptive devices (or does not require wheelchair)

17. Transfers: ground-wheelchair

- o Requires assistance
- o Transfers independent with or without adaptive devices (or does not require wheelchair)

18. Sexual Function

Sexual function is very important to most victims of spinal injury. Please tick those that apply to you.

Male questions:

- o I get normal erections
- o I only get erections when stimulated by touch
- o I can achieve orgasm

- I can ejaculate

Female questions:

- I can be aroused without touch
- I can only be aroused by touch
- I can achieve orgasm
- I can feel menstruation

Oswestry Back Pain Disability Questionnaire

Please tick the best option in each section. These questions are about pain from your back.

Section 1 – Pain intensity

- I have no pain at the moment
- The pain is very mild at the moment
- The pain is moderate at the moment
- The pain is fairly severe at the moment
- The pain is very severe at the moment
- The pain is the worst imaginable at the moment

Section 2 – Personal care (washing, dressing etc)

- I can look after myself normally without causing extra pain
- I can look after myself normally but it causes extra pain
- It is painful to look after myself and I am slow and careful
- I need some help but manage most of my personal care
- I need help every day in most aspects of self-care
- I do not get dressed, I wash with difficulty and stay in bed

Section 3 – Lifting

- I can lift heavy weights without extra pain
- I can lift heavy weights but it gives extra pain
- Pain prevents me from lifting heavy weights off the floor, but I can manage if they are conveniently placed eg. on a table
- Pain prevents me from lifting heavy weights, but I can manage light to medium weights if they are conveniently positioned
- I can lift very light weights
- I cannot lift or carry anything at all

Section 4 – Walking

- Pain does not prevent me walking any distance
- Pain prevents me from walking more than 2 kilometres
- Pain prevents me from walking more than 1 kilometre

- Pain prevents me from walking more than 500 metres
- I can only walk using a stick or crutches
- I am in bed most of the time

Section 5 – Sitting

- I can sit in any chair as long as I like
- I can only sit in my favourite chair as long as I like
- Pain prevents me sitting more than one hour
- Pain prevents me from sitting more than 30 minutes
- Pain prevents me from sitting more than 10 minutes
- Pain prevents me from sitting at all

Section 6 – Standing

- I can stand as long as I want without extra pain
- I can stand as long as I want but it gives me extra pain
- Pain prevents me from standing for more than 1 hour
- Pain prevents me from standing for more than 3 minutes
- Pain prevents me from standing for more than 10 minutes
- Pain prevents me from standing at all
-

Section 7 – Sleeping

- My sleep is never disturbed by pain
- My sleep is occasionally disturbed by pain
- Because of pain I have less than 6 hours sleep
- Because of pain I have less than 4 hours sleep
- Because of pain I have less than 2 hours sleep
- Pain prevents me from sleeping at all

Section 8 – Sex life (if applicable)

- My sex life is normal and causes no extra pain
- My sex life is normal but causes some extra pain
- My sex life is nearly normal but is very painful
- My sex life is severely restricted by pain
- My sex life is nearly absent because of pain
- Pain prevents any sex life at all

Section 9 – Social life

- My social life is normal and gives me no extra pain
- My social life is normal but increases the degree of pain
- Pain has no significant effect on my social life apart from limiting my more energetic interests eg, sport
- Pain has restricted my social life and I do not go out as often
- Pain has restricted my social life to my home
- I have no social life because of pain

Section 10 – Travelling

- I can travel anywhere without pain
- I can travel anywhere but it gives me extra pain
- Pain is bad but I manage journeys over two hours
- Pain restricts me to journeys of less than one hour

- o Pain restricts me to short necessary journeys under 30 minutes
- o Pain prevents me from travelling except to receive treatment

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Figure 26

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Figure 37

Email from Martin Baker dated 7 April 1014

Sir

Thank you for your enquiry and yes, we would be happy to provide these images - if you don't need them as very large files, you can download them from our site.

We would be very interested to see a copy of your thesis when finished - would that be possible?

Kind regards

Sarah

-----Original Message-----
From: martinb@phoenix.webexpectations.net
[mailto:martinb@phoenix.webexpectations.net]
Sent: 03 April 2014 11:43
To: martinbakeraircraft@gmail.com; Sarah Jeffery
Subject: Martin-Baker Contact Form

Martin-Baker Contact Form
Name: Edward Spurrier
Email: e.spurrier13@imperial.ac.uk
Company: Centre for Blast Injury Studies, Imperial College London
Enquiry: Good morning. I'm writing an MD thesis on blast injury in the spine, using ejector seats as a comparator. I wonder if I could please request permission to use some images from your website in my thesis? Specifically, for now:
http://www.martin-baker.com/products/ejection-seats/mk14#prettyPhoto[gallery]/4/
and
http://www.martin-baker.com/products/ejection-seats/mk16/eurofighter#
Many thanks
Ed Spurrier
Sqn Ldr RAF

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