

1 **An Investigation into Axial Impacts of the Cervical Spine using Digital Image Correlation**

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16

17 **Keywords**

18

19 Impact; spine; injury; digital image correlation; DIC

20

1 **Abstract**

2

3 **Background Context:** High-energy impacts are commonly encountered during sports such as Rugby
4 Union. Whilst catastrophic injuries resulting from such impacts are rare, the consequences can be
5 devastating for all those involved. A greater level of understanding of cervical spine injury
6 mechanisms is required, with the ultimate aim of minimizing such injuries.

7 **Purpose:** The present study aimed to provide a greater understanding of cervical spine injury
8 mechanisms, by subjecting porcine spinal specimens to impact conditions based on those measured
9 in vivo. The impacts were investigated using high-speed digital image correlation (DIC), a method not
10 previously adopted for spinal impact research.

11 **Study Design:** In-vitro biomechanical study

12 **Methods:** The study was funded through an institutional grant from the Rugby Football Union
13 Injured Players Foundation. Eight porcine specimens were impacted using a custom-made rig. The
14 cranial and caudal axial loads were measured at 1 MHz. Video data were captured with two cameras
15 at 4 kHz, providing measurements of the 3D deformation and surface strain field of the specimens
16 using DIC.

17 **Results:** The injuries induced on the specimens were similar to those observed clinically. The mean
18 (\pm SD) peak caudal load was 6.0 (\pm 2.1) kN, which occurred 5.6 (\pm 1.1) ms after impact. Damage
19 observable with the video data occurred in six specimens, 5.4 (\pm 1.1) ms after impact, and the peak
20 surface strain at fracture initiation was 4.6 (\pm 0.5) %.

21 **Conclusions:** This study has provided an unprecedented insight into the injury mechanisms of the
22 cervical spine during impact loading. The posture represents a key factor in injury initiation, with
23 lordosis of the spine increasing the likelihood of injury.

24

1 **Introduction**

2

3 Sports such as Rugby Union (rugby) routinely involve high-energy impact forces. On rare occasions,
4 these impacts can result in catastrophic injuries to the cervical spine that have devastating
5 consequences for all those involved. Whilst the risk of injury or fatality in rugby is comparable to
6 similar activities involving contacts, and to general employment-related incidents [1], it remains
7 important to minimize such risks. It has been estimated that approximately 40 % of serious cervical
8 spine injuries in rugby occur during scrummaging [2], and the relatively controlled nature of the set-
9 scrum (in which the eight forward players of each team bind together and then engage with one
10 another under the referee’s instruction) may offer the best potential to reduce risk through an
11 improved understanding of injury mechanisms, and appropriate guidelines to the governing bodies
12 of the game.

13

14 Previous research investigating axial impacts and injury mechanisms in the cervical spine has focused
15 on head constraint [3], the effect of the impact surface angle and padding [4, 5], vertebral
16 compression [6], horizontally positioned impacts [7], the effect of axial preload [8], and burst
17 fracture injuries [9]. These studies have demonstrated the complex nature of spinal injuries, and the
18 variation of injury mechanism(s) under seemingly similar conditions. Assessments of the injury
19 mechanisms due to axial loading encountered in rugby scrummaging, have been reported as being
20 due to hyperflexion, and buckling [2, 10, 11]. However, care must be taken in classifying cervical
21 spine injury mechanisms, as injury may have occurred due to more than one mechanism, and prior
22 to any outwardly observable change in head position [3, 12]. It may, therefore, be necessary to focus
23 on specific forms of impact situations in a highly controlled manner, in order to fully understand the
24 likely injury mechanisms, and how they may be avoided.

25

1 Catastrophic neck injury during scrummaging may occur as a result of a head impact to a front row
2 player due to improper engagement between the front rows of the opposing teams, or during the
3 collapse of a scrum [2]. Should a head impact occur, severe pocketing will result as a player's head
4 lodges in the shoulder and neck areas of the front row of the opposing team. Such conditions give
5 rise to a cervical spine impact in which both the cranial and caudal regions are highly constrained.

6

7 The aim of the present study was to provide an increased understanding of the injury mechanisms
8 associated with constrained axial impacts of the cervical spine. The testing conditions for the
9 investigation were aimed at replicating the constraints of rugby scrummaging. Injury mechanisms
10 were investigated using a combination of load measurement and high-speed imaging. Digital image
11 correlation (DIC), which has not previously been used in spinal impact research, was used to
12 investigate both movements and strain fields induced by the impact on a multi-level cervical spine
13 specimen.

14

15 **Methods**

16

17 Eight porcine cervical spinal specimens (C2-C6) were harvested from pigs aged between eight and
18 twelve months at the time of slaughter, with an average mass of approximately 60 kg (Bartlett &
19 Sons Ltd., Bath, UK). The specimens were dissected from longer sections of spine (C1-T2), all
20 musculature was removed, but facet capsules and ligaments were left intact with the exception of
21 the anterior longitudinal ligament (ALL). The ALL was removed to provide better visibility to the
22 anterior aspect of the vertebral bodies, which was used to measure the surface strain field using DIC.
23 Pilot holes were made in the lateral aspects of the vertebral bodies to accommodate steel eyelets
24 for the application of the follower-load. Following the dissection, specimens were wrapped in paper
25 towel, sprayed with 0.9 % saline solution, triple-sealed in plastic bags, and then stored at -24 °C.

26

1 Prior to impact testing, μ CT scans were acquired for each specimen using a Nikon XT225 ST (Nikon
2 Metrology UK, Hertfordshire, UK), and a Perkin Elmer PE1620 detector (PerkinElmer,
3 Buckinghamshire, UK). A total of 2,160 projections were acquired for each specimen. Image
4 reconstruction was performed with CT Pro 3D software (Version 3.1.3, Nikon Metrology), resulting in
5 a voxel resolution of 0.06 to 0.10 mm.

6

7 On the morning of testing each specimen was left to thaw for six hours at room temperature
8 (20 ± 2 °C) whilst still triple-sealed in plastic bags. During the last hour of thawing, the specimen was
9 removed from the plastic bags, eyelets were screwed into the vertebral bodies, and three self-
10 tapping screws were driven into the cranial and caudal ends to aid stability when potted in using low
11 melting point alloy (MCP75; Mining & Chemical Products Ltd., Northamptonshire, UK). Care was
12 taken during potting to ensure that the C3-C4 disc was aligned with the horizontal plane, and the
13 specimen was in the neutral position.

14

15 Following potting, the posterior aspect of each specimen was covered with paper towel sprayed with
16 0.9 % saline solution, while the anterior aspect of the vertebral bodies was dried with paper towel,
17 and painted with white paint, followed by a speckle spray of black paint to provide a means to
18 perform the DIC measurements. The specimen was then mounted in a custom-made impact fixture,
19 and a follower-load was applied to each side of the specimen using constant force springs rated at
20 51 N (CFS5.2, Misumi Europa GMBH, Schwalbach, Germany) via Bowden cables. This resulted in a
21 follower-load of 51 N on each side of the specimen, with a further 50 N applied to the cranial end of
22 the specimen due to the weight of the impact plate. This preload magnitude is comparable to
23 previous in-vitro cervical spine studies that have replicated the weight of the head and the stiffening
24 effect due to passive muscle activity [8, 13, 14].

25

1 The impact fixture (Figure 1) comprised an impact plate linked to a frame via two double linear
2 bearing units (Model LTDR25, AB SKF, SE-415 50 Göteborg, Sweden). The impact fixture held the
3 caudal end of the specimen rigidly, and allowed only vertical movement to the cranial end, which
4 aimed to replicate the constrained nature of an improper scrummage engagement in which the
5 torso is constrained, and the head severely pocketed.

6

7 Impacts were applied to the impact plate via a falling mass of 12.86 kg constrained with a custom
8 linear bearing assembly. A drop height of 250 mm was used to produce an impact velocity of
9 approximately 2.2 m/s, which is similar to relative impact velocity of scrum engagement measured
10 in-vivo using the “crouch, bind, set” call [15]. This call was introduced by the International Rugby
11 Board in 2014 (Law 20.19(g) Law Amendment Trial [16]). A layer of 5 mm thick nitrile rubber was
12 bonded to the top surface of the impact plate to prevent ringing vibrations. Two 22 kN compression-
13 extension load cells (Model SLC41/005000, RDP Electronics Ltd., Wolverhampton, UK) were used,
14 one mounted between the cranial specimen pot and the impact plate, and one mounted between
15 the caudal specimen pot and the baseplate. Load data was captured using a TiePie Handyscope HS5
16 (TiePie Engineering, Sneek, The Netherlands) and TiePie Multi Channel software (Version 1.0.29.0,
17 TiePie Engineering).

18

19 Image data was acquired with two high-speed cameras (Fastcam SA3 Master and Slave, Photron
20 Europe Ltd., West Wycombe, Buckinghamshire, UK) using Photron FASTCAM Viewer software (PFV
21 Version 3.3.5.0, Photron Europe Ltd.). Digital image correlation of the video data was completed
22 using Vic-3D (Version 2009.1.0, Correlated Solutions, Inc., Columbia, USA). The DIC system was
23 automatically calibrated prior to testing each specimen using fifty images of a 120 x 90 mm target
24 with 5 mm spacing.

25

1 Synchronized acquisition of the load and image data was triggered as the mass was dropped, with
2 load and image data acquired at 1 MHz and 4 kHz respectively.

3
4 After testing, 24 x-ray images were acquired of each specimen at intervals of 15° in the axial plane.
5 These images were used in conjunction with physical inspection to determine the damage induced.

7 **Data Analysis**

8
9 The load data was filtered with a fourth-order dual-pass, low-pass Butterworth filter with a cut-off
10 frequency of 5 kHz (Matlab Version R2012b, Mathworks, Inc., Natick, MA, USA). A number of
11 parameters of interest were calculated, the definitions of which are detailed below. For ease of
12 communication relevant parameters are reported in terms of mean (\pm standard deviation)
13 throughout this manuscript.

14
15 The impact time was identified as the point at which the load in the cranial load cell exceeded 200 N
16 [8]. The data period analyzed included the time of impact to 0.03 s post-impact; this encompassed
17 the entire primary impact for all specimens. The magnitudes of the maximum cranial (F_{cra}) and
18 caudal (F_{cau}) loads were identified, as was the time from impact until the maximum cranial (T_{cra}) and
19 caudal (T_{cau}) loads. The time to injury ($T_{2/3}$) was also identified, which was defined as the time at
20 which the caudal load dropped to 2/3 of the peak level [8]. The image data were used to identify the
21 time at which observable injury occurred (T_{inj}), defined as the frame at which fracture or major disc
22 pathology such as a prolapse, was induced. The peak cranial and caudal loads were compared using
23 Wilcoxon Signed Rank Tests. Comparisons were made of T_{cau} , T_{inj} , and $T_{2/3}$ using a Friedman's Test
24 with Wilcoxon Signed Rank Tests for post-hoc analysis. Significance was based on a p value of 0.05 in
25 all comparisons. Statistical analyses were performed with SPSS software (Version 21.0.0.0, IBM
26 Corporation, Armonk, NY, USA).

1 Images from 0.05 s prior to the impact to 0.03 s after the impact were analyzed using DIC (141 video
2 frames). For each specimen, the frame at 0.05 s prior to impact was used as the reference frame (at
3 which the strain was considered zero); measurements of the major principal strain magnitude, the
4 strain in the axial direction, and the 3D movement relative to the reference frame were assessed.

5

6 The DIC system allowed data to be obtained for each inspection point of an image for every frame of
7 the analysis. The arrangement in the present study resulted in the entire anterior aspect (both
8 vertebral bodies and intervertebral discs) of each specimen equating to approximately 3000
9 inspection points. This provided a large amount of data but artefacts around the edge of the analysis
10 area, and losses of information in certain frames, made automated data analysis unreliable.
11 Therefore, the DIC data were used to assess the behaviour of the specimens generally, and at
12 specific times and locations of interest.

13

14 **Results**

15

16 The pre-impact μ CT data and visual inspection showed that all specimens were healthy, with no
17 evidence of injury, disease, or degeneration. Impacts resulted in major damage to six of the eight
18 specimens.

19

20 The impact was rapidly transferred through the spine to the caudal load cell (Figure 2). The impact
21 was defined as the moment that the load exceeded 200 N in the cranial load cell; in all specimens
22 this load was reached at the caudal end in less than 1 ms ($0.8 (\pm 0.1)$ ms). The movement of the C5
23 level during the impact often blocked camera visibility of the C6 level; therefore, DIC analysis was
24 completed on the C2-C5 levels only.

25

1 The maximum load in the cranial and caudal load cells was 5.8 (± 2.0) kN and 6.0 (± 2.1) kN
2 respectively, which was reached at a time of 5.1 (± 1.0) ms and 5.6 (± 1.1) ms respectively after
3 impact (Table 1). Damage observable from the video data occurred in six out of eight specimens,
4 5.4 (± 1.1) ms after impact. The time to injury, $T_{2/3}$, as defined from the caudal load cell data, for the
5 six specimens in which observable injury occurred, was 7.9 (± 2.1) ms.

6

7 Damage included anterior fractures of the vertebral body (Figure 3), bilateral dislocation with facet
8 fractures, disc prolapses and/or decompressions, and fat emboli being ejected from the vertebral
9 bodies. All major injuries occurred in the C4-C6 region of the spine. Anterior inferior extension
10 teardrop fractures occurred at the C5 level in four specimens at the junction of the vertebral body
11 and the C5-C6 intervertebral disc. In one of these specimens the anterior inferior fracture was not
12 complete but was combined with vertical fracture of the anterior vertebral body in the sagittal plane
13 (Figure 4). One specimen sustained an incomplete anterior superior teardrop fracture to the C6 level
14 at the junction of the vertebral body and the C5-C6 disc. All disc prolapses were limited to the C5-C6
15 disc, though some decompression occurred at the C4-C5 disc in one specimen. Facet capsule injury
16 also occurred at the C5-C6 level, with the exception of one case where it was observed at the C4-C5
17 level. Facet fractures were observed at the C6 level, with one case at the C5 level.

18

19 The peak caudal load was significantly higher than the peak cranial load ($p=0.036$). Comparisons
20 between T_{cau} , $T_{2/3}$, and T_{inj} were only completed for the six specimens where injury was observed in
21 the high-speed images (CS1-2, CS4 and CS6-8). No significant difference was found between T_{cau} and
22 T_{inj} ($p=0.595$) but $T_{2/3}$ occurred significantly later than both T_{cau} ($p=0.028$) and T_{inj} ($p=0.028$).

23

24 The vertebral position in the sagittal, coronal, and axial planes were analyzed at 2.5 ms intervals
25 using single points taken at the centre of the anterior aspect of each vertebral body using a target-
26 based template. Sagittal displacement was greatest at the C4 level, and this occurred 7.5 ms after

1 impact, amounting to 4.75 (± 1.26) mm, with the two lowest values of 2.89 and 3.72 mm
2 corresponding to the uninjured specimens. The mean vertical displacement was greatest 10 ms after
3 impact at all vertebral levels with values of 5.45 (± 1.57) mm, 4.70 (± 1.32) mm, 4.80 (± 1.20) mm, and
4 5.29 (± 1.27) mm for C2, C3, C4, and C5 respectively. The displacement in the coronal plane was
5 below 1.50 mm in all specimens at all vertebral levels during the impacts.

6

7 The 3D motion analysis demonstrated that all specimens underwent increased lordosis due to the
8 impact resulting in a tendency to first order buckling (C-shaped) in the caudal region of the spine.
9 This was more pronounced in those specimens that suffered injuries but similar observations were
10 made when no injury was induced.

11

12 The major principal surface strain in the vertebral bodies reached values in excess of 4 % prior to
13 fracture, with mean peak strains of 4.6 (± 0.5) %. The maximum principal strains occurred in the axial
14 direction, and the anterior body fractures occurred as a result of tension. The axial compression due
15 to the impact exhibited lower strain values, with a mean peak compressive strain in the C2 vertebrae
16 of 2.6 (± 1.4) %, which occurred 0.6 (± 0.5) ms after the peak caudal load was reached.

17

18 **Discussion**

19

20 This study aimed to provide a greater understanding of the injury mechanisms of the cervical spine
21 due to axial impacts relating to misdirected rugby scrummaging. The constraint and impact velocity
22 of impacts in vivo were simulated in vitro using porcine cervical spine specimens. Specimens were
23 tested in a neutral posture with a physiological preload, and resulted in structural damage
24 comparable to axial impacts in vivo. The loads transferred through the spine were measured, and
25 high-speed imaging used with DIC to measure the surface strain field of the anterior aspect of the

1 vertebral bodies, and provide 3D reconstructions of the vertebral movement relative to the pre-
2 impact position.

3
4 Porcine specimens were used, as they provide similar characteristics to human cadaveric specimens
5 [17, 18]. Previous impact studies with human cadaveric specimens have used specimens with a
6 mean age from 52 years [3] up to 87 years [7]. Specimens toward the upper end of this range would
7 be more likely to sustain fractures, and soft tissue injury due to axial impacts than a regular rugby
8 player in their twenties or thirties. Such differences may lead to inconsistencies between the failure
9 mechanisms observed in-vitro compared to those sustained in young, healthy adults in-vivo. Porcine
10 specimens provide a consistent means to test spinal specimens with good bone density, and without
11 disc degeneration, which may reasonably reflect the skeletal profile of a young adult rugby player.

12
13 Previous studies have reported peak neck forces of approximately 2 kN as a result of peak head
14 forces of 3 to 12 kN [3-5, 8]. The present study used an impact velocity lower than many previous
15 studies in order to better represent velocities measured in vivo for scrummaging [15]. The drop mass
16 was also lower (12.86 kg) than previously used [3, 4, 8]. However, the peak caudal loads were higher
17 than most previously reported values, though they were within the range of cranial and caudal loads
18 measured in-vitro by Ivancic [7] of 7.5 (± 0.76) kN, and similar to the engagement forces of machine-
19 based scrummaging measured in vivo using a “fold-in” technique, which ranged from 4.25 (± 0.8) kN
20 in high-school teams to 8.6 (± 2.0) kN in International teams [19].

21
22 A physiological follower-load was used to simulate the passive muscle activity required to stabilize
23 the cervical spine, and specimens were constrained such that the most caudal vertebra was rigidly
24 fixed (C6), and the most cranial vertebra (C2) was constrained so as to only move axially. This
25 combination of preload application and constraint made the specimens less prone to buckling, and

1 therefore able to transfer greater axial load through the spine, including two specimens that
2 sustained peak loads of over 6 kN without major injuries.

3

4 Saari et al. [8] reported a mean time to injury of 2 ms with a follower-load, which is considerably less
5 than the mean time to observed injury of 5.9 ms in the present study. Mean time to injury without a
6 follower-load or padded impact surface has been reported as 6 ms [8], and 5 ms [3]. This suggests
7 that the results of the present study fit within the range of published data, but the differences
8 between studies may relate to both the specimen constraint and the application of a physiological
9 preload. This highlights the importance of testing specimens in a manner that is representative of
10 the in vivo scenario. Furthermore, it has been recently shown that muscle activity prior to
11 scrummage engagement differs according to the engagement technique [20]. The pre-activation of
12 muscles in the neck and torso would provide greater stability to the spine, and should be considered
13 in the design of future impact studies.

14

15 It is noteworthy that $T_{2/3}$ occurred significantly later than observed injuries. The time of observed
16 injury, T_{inj} , did not relate to specific characteristics of the load data, but no significant difference was
17 found between this time and the time of peak caudal load, T_{cau} . Therefore, the time of peak load
18 combined with high-speed video data is a more reliable way to assess time to injury than a defined
19 drop-off magnitude following peak caudal load.

20

21 The strain data for the anterior aspect of the vertebral bodies prior to and at the point of fracture
22 was comparable with previously reported values of the ultimate tensile strain of cortical bone in
23 bovine femurs [21], and slightly higher than the published data for the ultimate tensile strain of
24 trabecular bone from human cadaveric vertebrae [22].

25

1 Damage in the specimens of the present study was focused around the C5 and C6 levels, which
2 corresponded to the principal area over which the lordotic curve of the cervical spine occurred. Such
3 injuries reflect those seen clinically, with most cervical spine injuries due to axial impact occurring in
4 the caudal region of the cervical spine [10].

5

6 The extension-compression injuries of the present study are similar to previous in vitro tests [8], and
7 the anterior inferior fractures observed have also been reported in approximately one third of 156
8 cervical spine injuries in American Football players due to what was thought to be axial loading alone
9 [23]. Additionally, the dislocations combined with facet fractures observed in the present study have
10 also been reported in vivo [2, 10, 11].

11

12 The present study has shown that the above injuries can occur as a result of axial impacts but
13 demonstrated from the DIC measurements that anterior fractures resulted from tension in the
14 vertebral bodies due to first order buckling of the cervical spine in extension. Such loading conditions
15 are likely to be similar to those resulting from an improper engagement during scrummaging, with a
16 large axial load transferring down through a severely pocketed head to a highly constrained torso.
17 However, small changes in alignment and constraint could considerably alter the load transfer, and
18 therefore the injury mechanisms. Less constraint would provide the potential to escape from a direct
19 axial impact. Alternatively, a straighter cervical spine in similarly constrained conditions may be
20 better able to sustain higher loads; this may reduce the likelihood of buckling, but increase the
21 likelihood of burst fractures in the caudal region of the cervical spine.

22

23 The study has demonstrated the viability of using DIC for the 3D analysis and surface strain
24 measurement of spinal specimens undergoing high-speed impacts. The data compares well with
25 previous studies but also highlights the need for caution in interpreting the results from specific
26 impact studies to various in vivo scenarios. Whilst impact conditions may appear similar, the load

1 transfer and injury mechanisms of the spine are complex; small changes in posture and constraint
2 can lead to large differences in the magnitude and timing of impact forces, and the failure
3 mechanisms of the spine. Whilst the tests of the present study were based on potential impacts
4 during rugby scrummaging, the findings can be transferred to other scenarios involving constrained
5 axial impacts to the cervical spine, and may therefore be applicable to sports such as American
6 Football.

7

8 Further research is required to combine in vivo data of the posture, constraint, and muscle forces
9 under normal scrummaging conditions, to more closely simulate impacts in vitro. This will provide a
10 greater understanding of cervical spine injury mechanisms, and how the risk of them occurring
11 during sports such as rugby union can be minimized.

12

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14

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17

1 **References**

2

3 1. Fuller CW. Catastrophic Injury in Rugby Union Is the Level of Risk Acceptable? Sports
4 Medicine. 2008;38(12):975-86.

5 2. Trewartha G, Preatoni E, England ME, Stokes KA. Injury and biomechanical perspectives on the
6 rugby scrum: a review of the literature. British Journal of Sports Medicine. 2014.

7 3. Nightingale RW, McElhaney JH, Richardson WJ, Myers BS. Dynamic responses of the head and
8 cervical spine to axial impact loading. Journal of Biomechanics. 1996;29(3):307-18.

9 4. Nightingale RW, McElhaney JH, Richardson WJ, Best TM, Myers BS. Experimental impact injury
10 to the cervical spine: Relating motion of the head and the mechanism of injury. Journal of
11 Bone and Joint Surgery American Volume. 1996;78A(3):412-21.

12 5. Nightingale RW, Richardson WJ, Myers BS. The effects of padded surfaces on the risk for
13 cervical spine injury. Spine. 1997;22(20):2380-7.

14 6. Yoganandan N, Pintar FA, Sances A, Jr., Reinartz J, Larson SJ. Strength and kinematic response
15 of dynamic cervical spine injuries. Spine. 1991;16(10 Suppl):S511-7.

16 7. Ivancic PC. Biomechanics of sports-induced axial-compression injuries of the neck. Journal of
17 athletic training. 2012;47(5):489-97.

18 8. Saari A, Dennison CR, Zhu Q, et al. Compressive follower load influences cervical spine
19 kinematics and kinetics during simulated head-first impact in an in vitro model. Journal of
20 Biomechanical Engineering. 2013;135(11):111003.

21 9. Wilcox RK, Allen DJ, Hall RM, Limb D, Barton DC, Dickson RA. A dynamic investigation of the
22 burst fracture process using a combined experimental and finite element approach. European
23 Spine Journal. 2004;13(6):481-8.

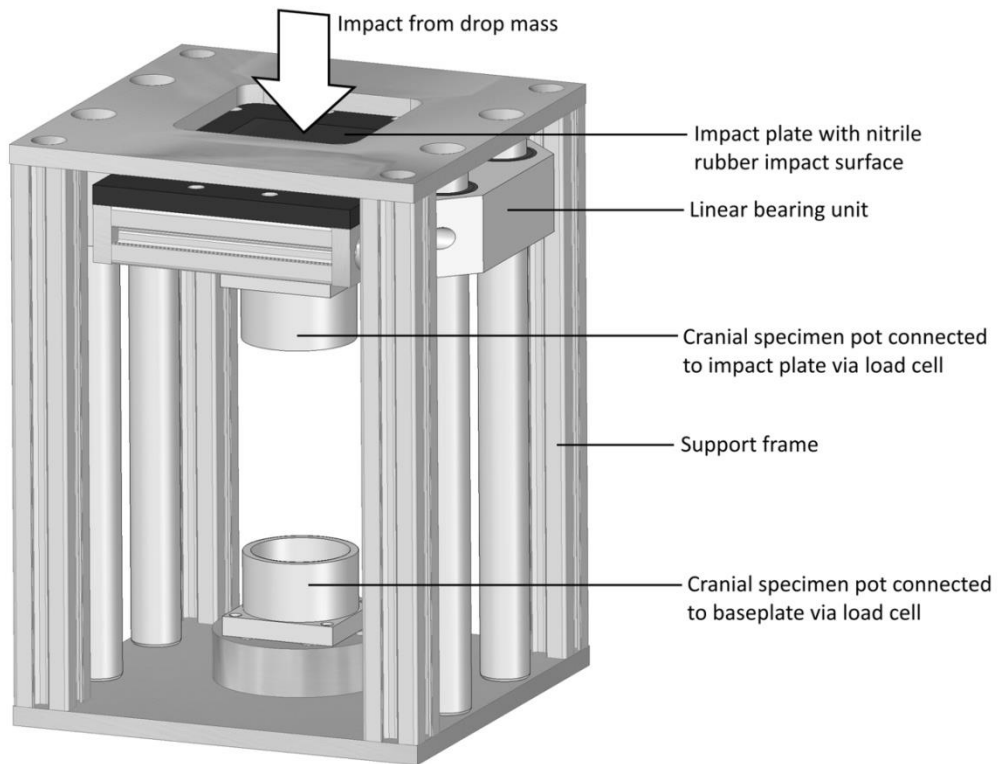
24 10. Kuster D, Gibson A, Abboud R, Drew T. Mechanisms of cervical spine injury in rugby union: a
25 systematic review of the literature. British Journal of Sports Medicine. 2012;46(8):550-4.

- 1 11. Dennison CR, Macri EM, Crompton PA. Mechanisms of cervical spine injury in rugby union: is it
2 premature to abandon hyperflexion as the main mechanism underpinning injury? *British*
3 *Journal of Sports Medicine*. 2012;46(8):545-9.
- 4 12. Winkelstein BA, Myers BS. The biomechanics of cervical spine injury and implications for injury
5 prevention. *Medicine and Science in Sports and Exercise*. 1997;29(7):S246-S55.
- 6 13. Miura T, Panjabi MM, Crompton PA. A method to simulate in vivo cervical spine kinematics using
7 in vitro compressive preload. *Spine*. 2002;27(1):43-8.
- 8 14. Ivancic PC, Panjabi MM, Ito S, Crompton PA, Wang JL. Biofidelic whole cervical spine model with
9 muscle force replication for whiplash simulation. *European Spine Journal*. 2005;14(4):346-55.
- 10 15. Cazzola D, Preatoni E, Stokes KA, England ME, Trewartha G. A modified prebind engagement
11 process reduces biomechanical loading on front row players during scrummaging: a cross-
12 sectional study of 11 elite teams. *British Journal of Sports Medicine*. 2014.
- 13 16. International Rugby Board. *Rugby Union Laws of the Game*. 2014 ed. Dublin, Ireland:
14 International Rugby Board; 2014.
- 15 17. Wilke H-J, Geppert J, Kienle A. Biomechanical in vitro evaluation of the complete porcine spine
16 in comparison with data of the human spine. *European Spine Journal*. 2011;20(11):1859-68.
- 17 18. Busscher I, Ploegmakers J, Verkerke G, Veldhuizen A. Comparative anatomical dimensions of
18 the complete human and porcine spine. *European Spine Journal*. 2010;19(7):1104-14.
- 19 19. Preatoni E, Stokes KA, England ME, Trewartha G. Engagement techniques and playing level
20 impact the biomechanical demands on rugby forwards during machine-based scrummaging.
21 *British Journal of Sports Medicine*. 2014.
- 22 20. Stone B, Cazzola D, McGuigan P, et al., editors. *Spinal muscle activity in simulated rugby union*
23 *scrummaging is affected by different engagement conditions*. 10th Bath Biomechanics
24 Symposium; 15th September 2014; Bath, UK.
- 25 21. Kopperdahl DL, Keaveny TM. Yield strain behavior of trabecular bone. *Journal of*
26 *Biomechanics*. 1998;31(7):601-8.

- 1 22. Li S, Demirci E, Silberschmidt VV. Variability and anisotropy of mechanical behavior of cortical
2 bone in tension and compression. *Journal of the Mechanical Behavior of Biomedical Materials*.
3 2013;21(0):109-20.
- 4 23. Torg JS, Pavlov H, Oneill MJ, Nichols CE, Sennett B. The Axial Load Teardrop Fracture - a
5 Biomechanical, Clinical, and Roentgenographic Analysis. *American Journal of Sport Medicine*.
6 1991;19(4):355-64.
- 7

1 **Figures**

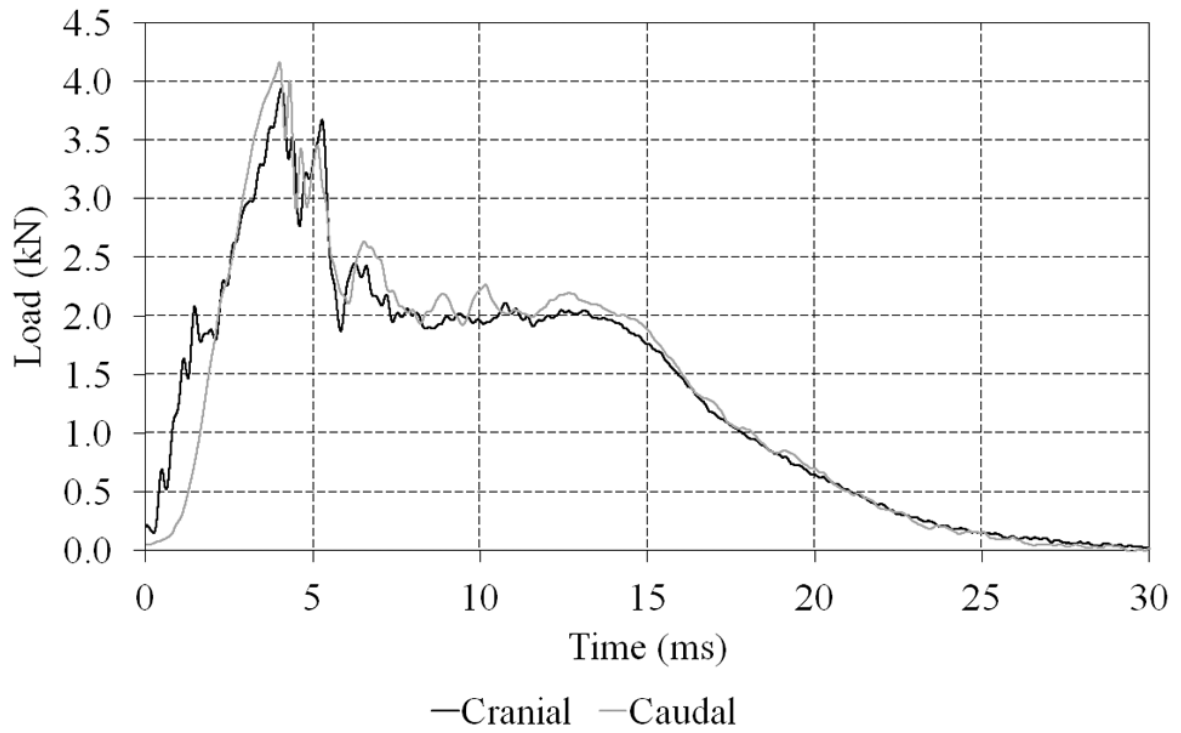
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3

4 Figure 1: Impact fixture used to constrain spinal specimens for axial impacts

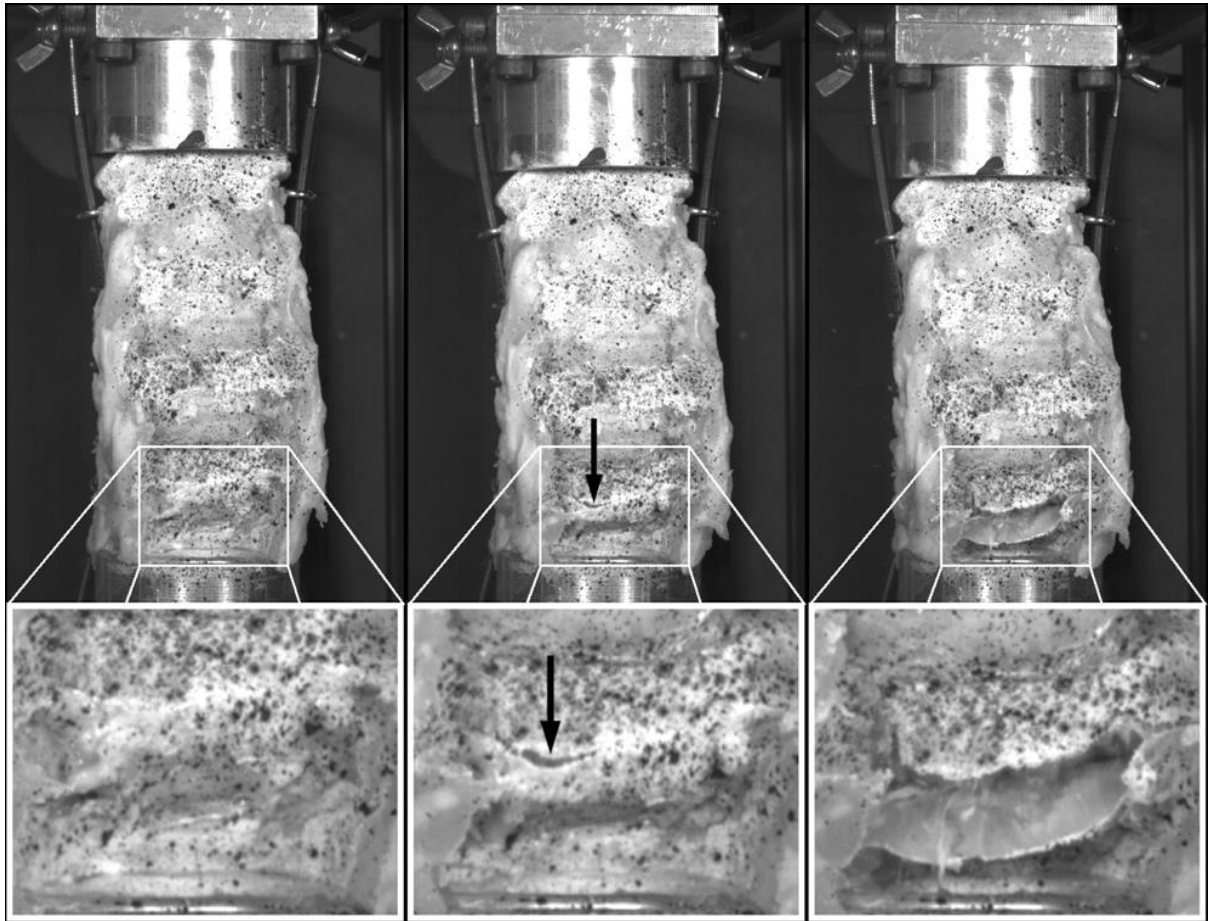
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2 Figure 2: Representative impact load profile taken from specimen CS1

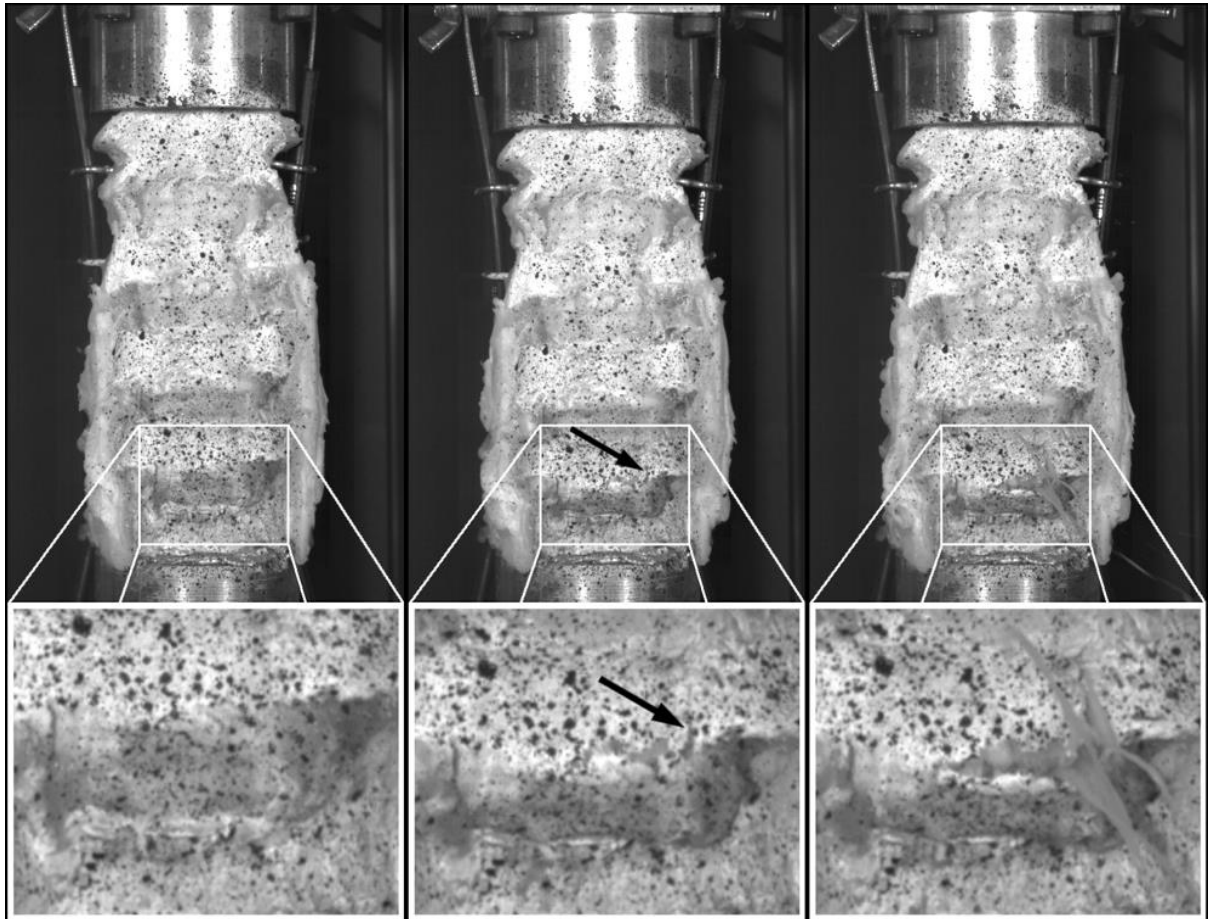
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1

2 Figure 3: Image frames of specimen CS7. Detail of the specimen at the time of impact (left). Crack
3 propagating at the C5 level at the time of peak caudal load, 5.8 ms after impact (centre) and anterior
4 vertebral fracture 10 ms after impact (right)

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2 Figure 4: Image frames of specimen CS8. Detail of the specimen at the time of impact (left). Crack
3 propagating vertically at the C5 level at the time of peak caudal load, 6.7 ms after impact (centre)
4 and fluid ejection 10 ms after impact (right)

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1 **Tables**

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3 Table 1: Load and time data for the eight porcine cervical specimens

Specimen	Peak Cranial	Peak Caudal	Time to	Time to	Time to	Time to 2/3
	Load (kN)	Load (kN)	Peak Cranial	Peak Caudal	observed	Peak Caudal
	F _{cran}	F _{caud}	Load (ms)	Load (ms)	injury (ms)	Load (ms)
	T _{cran}	T _{caud}	T _{inj}	T _{2/3}		
CS1	3.94	4.16	4.1	4.0	4.5	5.5
CS2	4.61	4.56	5.0	5.3	4.1	5.7
CS3	6.16	6.15	4.0	4.0	-	8.8
CS4	10.29	10.88	4.6	5.5	6.7	11.5
CS5	6.23	6.55	6.8	7.2	-	9.5
CS6	4.82	5.09	6.2	6.0	6.5	7.6
CS7	4.70	4.86	5.3	5.8	4.6	6.5
CS8	5.99	6.13	5.1	6.7	5.8	7.9
Mean	5.84	6.05	5.1	5.6	5.4	7.9*
SD	1.98	2.13	1.0	1.1	1.1	2.0*

4 * T_{2/3} calculated for only those specimens with injury (i.e. CS1, 2, 4, 6-8) was 7.4 (±2.2) ms