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Quantitative evaluation of facet deflection, stiffness, strain and failure load during simulated cervical spine trauma

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ABSTRACT

Traumatic cervical facet dislocation (CFD) is often associated with devastating spinal cord injury. Facet fractures commonly occur during CFD, yet quantitative measures of facet deflection, strain, stiffness and failure load have not been reported. The aim of this study was to determine the mechanical response of the subaxial cervical facets when loaded in directions thought to be associated with traumatic bilateral CFD – anterior shear and flexion. Thirty-one functional spinal units ($6 \times$ C2/3, C3/4, C4/5, and C6/7, $7 \times$ C5/6) were dissected from fourteen human cadaver cervical spines (mean donor age 69 years, range 48–92; eight male). Loading was applied to the inferior facets of the inferior vertebra to simulate the *in vivo* inter-facet loading experienced during supraphysiologic anterior shear and flexion motion. Specimens were subjected to three cycles of sub-failure loading (10–100 N, 1 mm/s) in each direction, before being failed in a randomly assigned direction (10 mm/s). Facet deflection, surface strains, stiffness, and failure load were measured. Linear mixed-effects models ($\alpha = 0.05$; random effect of cadaver) accounted for variations in specimen geometry and bone density. Specimen-specific parameters were significantly associated with most outcome measures. Facet stiffness and failure load were significantly greater in the simulated flexion loading direction, and deflection and surface strains were higher in anterior shear at the non-destructive analysis point (47 N applied load). The sub-failure strains and stiffness responses differed between the upper and lower subaxial cervical regions. Failure occurred through the facet tip during anterior shear loading, while failure through the pedicles was most common in flexion.

1. Introduction

Traumatic cervical facet dislocation (CFD) is often associated with devastating spinal cord injury, resulting in tetraplegia in up to 87% of cases (Hadley et al., 1992; Payer and Schmidt, 2005). CFD may be unilateral or bilateral, with bilateral facet dislocation (BFD) more often resulting in complete spinal cord injury (Allen et al., 1982; Quarrington et al., 2017). These injuries occur most commonly, and are most often survived, in the sub-axial region (C3-T1). They are frequently a result of traffic and sporting accidents, and falls (Allen et al., 1982; Quarrington et al., 2017), during which the external loading applied to the neck can be complex and variable.

BFD is thought to result from a *global, supra-physiologic* flexion moment about the subaxial cervical spine, caused by axial compressive forces applied to the head with large anterior eccentricity (Allen et al., 1982; Cusick and Yoganandan, 2002; Huelke and Nusholtz, 1986; White and Panjabi, 1990), or from inertial motion of the head during high deceleration events (Huelke and Nusholtz, 1986). In head-first impact tests of head-neck specimens, BFDs occurring in the lower cervical spine have been associated with local intervertebral flexion and anterior shear motions (Hodgson and Thomas, 1980; Ivancic, 2012; Nightingale et al., 2016).

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The inertial injury mechanism of BFD was validated in one experimental series (Ivancic et al., 2007, 2008; Panjabi et al., 2007) in which incrementally increasing, sagittal decelerations were applied to cervical motion segments (with a head mass surrogate) until dislocation occurred. Large flexion angles and anterior shear displacements were the dominant sagittal *intervertebral* motions observed during the injury event (Panjabi et al., 2007). Interestingly, no cervical facet fracture-dislocations have been produced experimentally, yet facet fractures are associated with up to 88% of clinical CFD cases (Foster et al., 2012). It has been suggested that concomitant fracture may be due to a large component of anterior shear in the local injury vector (Foster et al., 2012), but this has not been validated experimentally.

Studies that investigated the kinematics of cervical vertebrae during dynamic spinal motion have assumed that the anterior and posterior anatomy act as a rigid body (Ivancic et al., 2007, 2008; Panjabi et al., 2007). However, the high incidence of facet fracture associated with CFD would suggest that large loads are transmitted through this joint during the injurious motions, and one could expect substantial bending of the facets to occur prior to mechanical failure. In addition, sagittal bending of the facets in excess of 14°, relative to the vertebral body, was observed in a *lumbar* specimen during replicated *physiological* intervertebral flexion (Green et al., 1994). The magnitude of facet deflection and the mechanical response of the sub-axial cervical facets during loading to simulate supra-physiologic anterior shear and flexion motions have not been reported.

The mechanical response of the cervical facet capsule during simulated trauma has been well characterized, particularly regarding soft-tissue strains during 'whiplash' events (Cholewicki et al., 1997; Panjabi et al., 1998; Siegmund et al., 2008; Siegmund et al., 2001); however, strain data is not available for the bony facet. Investigations of the load-bearing capacity (Hakim and King, 1976; King et al., 1975; Pollintine et al., 2004), failure mechanisms (Cyron et al., 1976), fatigue strength (Cyron and Hutton, 1978) and surface strain response (Schulitz and Niethard, 1980; Shah et al., 1978; Suezawa et al., 1980) of the *lumbar* facets and neural arch have been performed, but similar analyses have not been reported for the subaxial cervical spine, or during simulated facet dislocation. Quantitative measures of the mechanical response of the cervical facets to simulated traumatic loading may be important for validation of computational models of cervical trauma and to inform design of advanced anthropometric test device (ATD) necks and associated injury criteria.

The aim of this study was to quantify the sagittal deflection, apparent stiffness, surface strain and failure load of subaxial cervical inferior facets under loads simulating the proposed injury vectors of supraphysiologic *in vivo* flexion and anterior shear motions.

2. Methods

2.1. Specimen preparation

Thirty-one functional spinal units (FSUs); six C2/3, six C3/4, six C4/5, seven C5/C6 and six C6/C7, were dissected from fourteen fresh-frozen human cadaver cervical spines (mean donor age 69 years, range 48–92; eight male). Radiographs and high-resolution computed tomography (CT) scans (Toshiba Aquilion ONE, Otawara, Japan; 0.5 mm slice thickness, 0.3 mm in-plane resolution) were obtained and each specimen was screened for excessive degeneration, injury and disease by a senior spinal surgeon. Average volumetric bone mineral density (vBMD) was quantified from CT using a calibration phantom (Mindways Software Inc., Texas, USA) and 'FIJI' image analysis software (1.51p, ImageJ, Maryland, USA) (Schindelin et al., 2012) (Fig. S1a). Vertebral endplate depths and sagittal facet angles were measured using FIJI (Fig. S1b and S1c).

Specimen musculature was removed and the vertebral disc and bilateral facet joint capsules were preserved (Fig. 1a). The vertebral bodies of each FSU were embedded in polymethylmethacrylate (PMMA; Vertex Dental, Utrecht, Netherlands) using a custom adjustable mold (Fig. 1b). To assist with fixation a wood screw was inserted through the vertebral bodies and disc, and steel wire was wrapped around the vertebral bodies and through the transverse foramen (Fig. 1a); excess wire and the screw-tip protruded from the superior endplate of the superior vertebra into a rectangular embedding cavity approximately 50 mm in length. The FSU was placed in the mold which was then filled with PMMA. A support bar was positioned within the spinal canal along the posterior surfaces of the vertebral bodies and was fixed to the PMMA block (Fig. 1b and c). Three types of support bars, accommodating variation in specimen geometry, were used to prevent embedding failure: (1) 90 × 20 × 1.5 mm aluminum; (2) 90 × 20 × 5 mm steel; and, (3) 90 × 10 × 5 mm steel.

2.2. Mechanical loading

Each specimen-PMMA assembly was rigidly mounted to the base of a biaxial materials testing machine (8874, Instron, High Wycombe, UK) via a custom support apparatus attached to a rotary table (VU150, Vertex, Taichung City, Taiwan) (Fig. 2). Using the rotary table, the inferior articular facet surfaces of the inferior vertebrae were positioned relative to the actuator to simulate the loading vectors thought to be applied by the opposing facets during *in vivo*, supraphysiologic flexion and anterior shear motions (Fig. 2). A 10 N pre-load and then three cycles of sub-failure loading to 100 N (a non-destructive load determined from pilot testing) were applied bilaterally to the geometric center of each articular facet surface at 1 mm/s using 6 mm diameter hemispherical

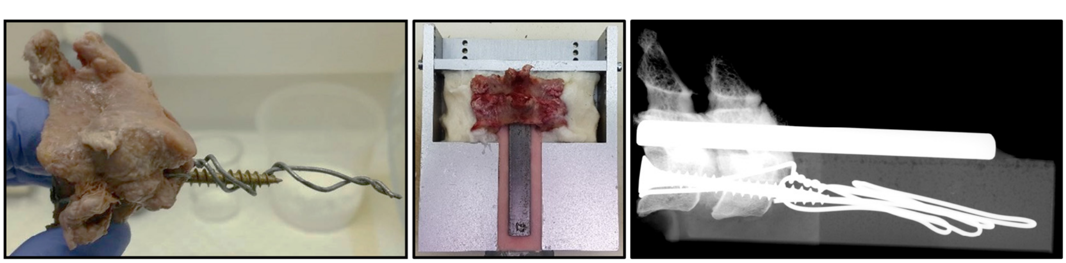
a b c

Fig. 1. Specimen preparation: (a) cervical functional spinal unit dissected of soft-tissue, with wood-screw and steel wire attached to the vertebral bodies. (b) The specimen was positioned in a custom mold with the spinous processes pointing vertically, perpendicular to the base, such that the posterior surfaces of the vertebral bodies aligned with the top surface. The lateral anatomy was pressed into plasticine to hold the specimen in the desired orientation, and to prevent the facets being embedded. The mold was then filled with PMMA and a support bar was fixed to the posterior surfaces of the vertebral bodies. (c) A lateral radiograph of the embedded specimen.