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Development of an equation for calculating vertebral shear failure tolerance without destructive mechanical testing using iterative linear regression

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ABSTRACT

Equations used to determine vertebral failure tolerances without the need for destructive testing are useful for scaling applied sub-maximal forces during *in vitro* repetitive loading studies. However, existing equations that use vertebral bone density and morphology for calculating compressive failure tolerance are unsuitable for calculating vertebral shear failure tolerance since the primary site of failure is the pars interarticularis and not the vertebral body. Therefore, this investigation developed new equations for non-destructively determining vertebral shear failure tolerance from morphological and/or bone density measures. Shear failure was induced in 40 porcine cervical vertebral joints (20 C3-C4 and 20 C5-C6) by applying a constant posterior displacement to the caudal vertebra at 0.15 mm/s. Prior to destructive testing, morphology and bone density of the posterior elements were made with digital calipers, X-rays, and peripheral quantitative computed tomography. Iterative linear regression identified mathematical relationships between shear failure tolerance, and morphological and bone density measurements. Along with vertebral level, pars interarticularis length and lamina height from the cranial vertebra, and inferior facet height from the caudal vertebra collectively explained 61.8% of shear failure tolerance variance. Accuracy for this relationship, estimated using the same group of specimens, was 211.9 N or 9.8% of the measured shear failure tolerance.

1. Introduction

In vitro testing of vertebrae and vertebral joints is important for quantifying biomechanical failure tolerances, establishing ergonomic limits, and identifying tissue injury mechanisms [1–8]. A recent review on occupational spine biomechanics emphasized the need for further *in vitro* investigations to identify acute tissue damage thresholds under modes of loading other than compression (e.g. shear), and to establish thresholds for cumulative spine loading [9]. In response to the second part of Potvin's [9] recommendation, the current study developed an equation that could be used to determine the shear failure tolerance of porcine cervical vertebral joints without the need for destructive testing. The equation developed from this study would provide a preliminary, but pivotal step for future *in vitro* studies using repetitive shear loading paradigms to establish the vertebral joint's cumulative shear tolerance, and the influence of sub-maximal shear force magnitude on the cumulative tolerance.

To investigate the influence of applied force magnitude on fatigue life in a repetitive loading protocol, it is often desirable to ensure that specimens can be grouped on the basis of acute injury potential imposed by repetitively applied sub-maximal forces [10]. As an example, these authors scaled repetitively applied sub-maximal compressive forces to a percentage of each specimen's estimated compressive failure tolerance to control the acute injury potential imposed by the applied force's magnitude. Normalizing the applied sub-maximal compressive force magnitude to a percentage of the predicted failure tolerance is also beneficial for enhancing comparisons of *in vitro* results between animal and human specimens. Specimen-specific estimates of acute compressive failure tolerance can be non-destructively obtained by using a linear regression model that mathematically relates measurements of the vertebral body's endplate area to acute vertebral joint compressive failure tolerances measured from *in vitro* tests [11]. This is a reasonable approach for predicting compressive failure tolerance since endplate fractures are commonly observed injuries resulting from compressive force during *in vitro* testing [12,13]. Other equations have also used measurements of the vertebral body's bone mineral density, either by itself or in conjunction with morphological measurements such as endplate area, to predict compressive strength [11,12,14–16]. In fact, combining measurements of vertebral morphology and bone density has been shown to improve

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predictions of compressive strength in human and macaque vertebral bodies [14,17], but did not enhance prediction of compressive strength in porcine cervical FSUs [11].

Given that vertebral tissue damage following destructive testing differs according to the mode of applied loading [3,5,13], it is unlikely that previously developed equations for determining the vertebral joint's compressive failure tolerance would also be suitable for determining failure tolerances in other modes of loading such as shear. Pars interarticularis (PI) fractures have been identified by *in vivo* and *in vitro* studies as the predominant injury associated with exposure to vertebral shear force [18,19]. Furthermore, these fractures are initiated at the caudal and ventral aspect of the pars [20]. Since the primary vertebral bony structures that interact under shear forces are the facets, a bending moment generated about the PI by facet articulation has been hypothesized as a shear injury mechanism for the spine [5,21,22]. Therefore, any mathematical model that attempts to determine the vertebral joint's acute shear failure tolerance without mechanical testing should likely include morphological measurements from the PI and/or facets and/or measurements of bone density. Likewise, other bony structures located posterior to the vertebral body, such as the lamina and pedicles have also been shown to influence facet interaction and vertebral mechanics with exposure to shear force [23–25]. Thus, it is possible that morphology of these structures may also be related to vertebral acute shear failure tolerance. Recent evidence using human lumbar specimens has shown that bone density may explain up to 59% of the variance in peak shear force when a FSU's cranial vertebra is displaced in an anterior direction relative to the stationary caudal vertebra [26]. Regression models that relate bone morphology and density of the elements located posterior to the vertebral body to shear failure tolerances may also help to identify critical parameters linked to fracture risk from facet interaction induced by shear loading.

This investigation developed new equations using stepwise linear regression that mathematically related vertebral morphological and/or bone density measurements to acute shear failure tolerances measured from *in vitro* tests using porcine cervical FSUs as surrogates for the human lumbar spine. Based on similar work performed for compressive loading of vertebrae [14,17], it was hypothesized that combining morphological and bone density measurements into a single equation would explain the most variance in measured acute vertebral shear failure tolerance.

2. Methods

2.1. Specimen preparation and assessment

2.1.1. Specimen assessment

Forty FSUs (20 C3-C4 and 20 C5-C6) were excised from twenty frozen porcine cervical spines obtained from a local abattoir. Each specimen was thawed overnight prior to removal of muscle and fat leaving an osteoligamentous FSU comprised of two vertebrae, the intervertebral disc, and ligaments. All specimens were classified, according to the grading system outlined by Galante [27] as non-degenerated (Grade 1) intervertebral discs by visual inspection of the exposed endplates, and were used in the study. Each exposed endplate's area was determined using the equation for area of an ellipse ($\pi WD/4$) with W being the mediolateral width and D being the anterior–posterior depth [28]. Average area between the two exposed endplates was used as an estimate of the FSU's intervertebral disc area [11]. Digital calipers (accuracy = ±0.02 mm) were used to measure height and width, and interfacet distances between the outside and inside edges of exposed superior and inferior facets respectively belonging to the FSU's cranial (either C3 or C5) and caudal (either C4 or C6) vertebrae. Pedicle and lamina height and width for both vertebrae were also measured with digital calipers. Surface areas for the exposed facets were also calculated using the equation for area of an ellipse. Facet angles, and the difference between left and right facet angles (tropism) were quantified from a transverse plane X-ray (ImageJ, National Institutes of Health, USA) [7,29].

2.1.2. Imaging of the pars interarticularis

For each FSU, six metal pins (diameter = 0.5 mm) were used to mark the superior and inferior border of the PI for each vertebra, and the caudal vertebra's inferior endplate and inferior facet tip (Fig. 1A). The superior PI border was defined as the lateral junction between the superior facet and the lamina while the inferior PI border was defined as the medial junction between the inferior facet and pedicle. One metal nail (diameter = 1.25 mm) marked a reference location on the cranial vertebra's spinous process.

A sagittal plane X-ray was taken of the specimen with the metal pins, nail, and a calibration frame. Due to irregularity of vertebral geometry, positioning for the X-ray was controlled to the unloaded and neutral posture by placing the specimen's right side onto a formed impression created in a 25.4 mm thick block of extruded polystyrene foam. Planar coordinates corresponding to each pin-head, nail and calibration frame endpoint were manually digitized (ImageJ, National Institutes of Health, USA) from the digitally developed X-ray (Fig. 1A). A line connecting the superior and inferior PI borders for each vertebra, which is consistent with fracture lines following shear failure [5,7,8] defined the PI (Fig. 1B). The length of this line defined the PI length. A second line connecting the most inferior points on the caudal vertebra's exposed inferior endplate and facet created a line that was coincident with the plane of the plastic plate on which the specimen was to be mounted for peripheral quantitative computed tomography (pQCT) scanning. This line defined the specimen's base. The acute angle between the lines that defined the PI and the specimen's base, along with the perpendicular distance between the digitized location for the nail and the PI were both calculated from the sagittal plane X-ray (Fig. 1C). These measurements were performed for both the cranial and caudal vertebrae of each specimen.

Metal pins were then removed from the specimen, but the nail remained as a reference point for pQCT scanning (XCT200L, Stratec Medizintechnik GmbH, Pforzheim, Germany). The specimen was wrapped in saline soaked gauze and cellophane to prevent specimen drying during scanning. Plastic zip ties rigidly affixed the specimen to a custom-built plastic jig with two plastic plates that were set at the aforementioned acute angle between the specimen's base and PI. Both specimen and jig were positioned between posts on either side of the pQCT's gantry so that the PI and scan planes were parallel. This was first performed for the FSU's cranial vertebra and then repeated for the caudal vertebra.

Before FSU scanning, a cone phantom was scanned to calibrate individual voxel intensities to measurements of density. For each specimen, five sequential images with a slice thickness of 1.1 mm, scan speed of 10 mm/s, and voxel size of 0.2 mm were acquired for both the cranial and caudal vertebrae's PI. The gantry's starting position (S) was determined using the following equation (1).

$$S = R + d - 2T \tag{1}$$

where R is the nail's reference position determined from an initial scout scan, d is the distance between the nail and the PI's superior border, and T is the slice thickness. Ensuring that separation between the nail and the PI's superior border was greater than twice the slice thickness eliminated beam hardening artifact in the third slice.