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# Biomechanical Response of the Human Clavicle: The Effects of Loading Direction on Bending Properties

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The purpose of this study was to determine the influence of loading direction on the structural response of the human clavicle subjected to three-point bending. A total of 20 clavicles were obtained from 10 unembalmed fresh-frozen postmortem human subjects ranging from 45 to 92 years of age. The right and left clavicles from each subject were randomly divided into two test groups. One group was impacted at 0° from the transverse plane, and the second group was impacted at 45° angle from the transverse plane. There was no statistically significant difference in peak force ($p = .22$), peak moment ($p = .30$), or peak displacement ($p = .44$) between specimens impacted at 0° versus 45° from the transverse plane. However, there was a significant difference in the structural stiffness ($p = .01$) and peak strain ($p < .01$) between specimens impacted at 0° versus 45° from the transverse plane. The peak strain, however, must be evaluated with caution because of the variation in fracture location relative to the strain gauge. Due to the controlled matched data set, the differences in the structural stiffness with respect to loading direction can be attributed to the complex geometry of the clavicle and not material differences.

***Keywords:*** failure, strength, stiffness, geometry, orientation

Clavicle fractures are classified as having an Abbreviated Injury Score (AIS) of 2, where AIS is a system used by researchers and trauma centers to rank and compare the severity of injuries on a scale of 1 (minor) to 6 (death) based on threat to life. However, clavicle fractures can result in either temporary or long-term loss of functional capacity. Housner and Kuhn (2003) reported that depending on the fracture type, clavicle fractures take 4–12 weeks to heal in adults. During this time, the range of motion and strength of the shoulder are limited. Midclavicle fractures are of special concern since they may lead to significant morbidity due to a high rate of nonunion, 22–33%, when treated nonoperatively (Nordqvist, 1993).

An analysis of the National Automotive Sampling System's Crashworthiness Data System (NASS-CDS) found that over 9,700 three-point belt-restrained occupants incur a clavicle fracture every year. For occupants exposed to frontal automotive impacts, over 90% of the fractures occurred in the clavicle as a result of being loaded by the shoulder belt, as evidenced by left clavicle fractures for drivers and the right clavicle fractures for right front-seat passengers. This injury pattern is consistent with the hypothesis that shoulder belt loading initiates clavicle fractures in frontal impacts. Therefore, understanding the biomechanics of clavicle fractures is an important factor in the optimization of seatbelt restraint systems.

There have only been two studies to the author's knowledge that have evaluated the biomechanical response of the human clavicle. Bolte et al. (2000) evaluated the strength of the clavicle by conducting three-point bending tests on six human clavicles at an impact rate of 0.5 mm/s. However, the direction of loading was not reported. Proubasta et al. (2002) conducted three-point bending tests on 5 intact clavicles dissected from fresh-frozen human cadavers. Before testing, the distal ends were fixed in aluminum cylinders using polyester resin. Proubasta et al. (2002) applied the load in the anterior-to-posterior direction 3 cm from the midpoint, medial or lateral not reported, of a fixed 150-mm test span at a loading rate of 0.05 mm/s.

Although there have been some studies that have investigated the biomechanical response of the clavicle in three-point bending, these studies have been limited to a single loading direction and low impact speeds. The clavicle has a complex shape, and differences found in the structural response of the clavicle due to different loading angles could provide insight on possible methods to optimize seatbelt restraint design and thus

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minimize the risk of clavicle fractures. Therefore, the purpose of this study was to evaluate the biomechanical response of the human clavicle when subjected to dynamic three-point bending in two loading directions, 0° and 45°, from the transverse plane.

## Methods

Dynamic three-point dynamic tests were performed at two loading angles on 20 human clavicles obtained from 10 unembalmed fresh-frozen postmortem human subjects ranging from 45 to 92 years of age (Table 1). Freezing was used as a means to preserve the specimens because numerous previous studies have indicated that freezing does not significantly affect the material properties of cortical bone, when frozen to a temperature of −20 °C (Sedlin, 1965; Weaver, 1966; Griffon et al., 1995; Linde & Sorensen, 1993; Frankel, 1960; Hamer et al., 1996). Numerous authors have reported significant differences in the material properties of dry bone compared with wet bone (Yamada, 1970; Dempster & Lippicoat, 1952; Evans & Lebow, 1951). Therefore, gauze pads soaked in normal saline were wrapped around the specimens and sealed in a plastic bag to maintain proper specimen hydration after the specimens were dissected from the bodies (Reilly et al., 1974; Burstein et al., 1974).

The right and left clavicles were randomly divided into two groups, and each group contained one specimen from each of the 10 matched pairs. Each group was subjected to an impact at either 0° or 45° from the transverse plane (Figure 1). Three-dimensional computed tomography (CT) scans of each clavicle were obtained before specimen preparation and testing to account for the irregular geometry of the clavicle cross section. An image processor was used to properly orient the three-dimensional CT reconstructions of the clavicles. Then the center of the clavicle corresponding to the point of load application in the experimental setup was located, based on the CT scans and experimental measurements, and a cross section was created. For the image analysis, the cross-sectional image was aligned such that the assumed loading direction was from the right (2). For specimens impacted 45° from the transverse plane, the image was rotated an additional 45°. The clavicle cross sections were then thresholded to obtain the cortical bone cross section, pixel value of 190/255, using the standard 8-bit black and white CT image imported into Adobe's Photoshop. The 190 pixel value resulted in an acceptable cortical bone cross section for all specimens, and preserved the variation between certain clavicles with very thin cross sections and other healthier bones with thicker cortex. Finally, each cross-sectional CT image was converted to binary black (0) and white (1), where pixels designated as 1 represented bone (Figure 3). It should be noted that the material properties of the cortical bone were assumed to be homogeneous.

A custom Matlab code was used to calculate the area moment of inertia (I), the distance from the neutral axis to the tensile surface (c), and the cortical bone cross-sectional area (A), using the thresholded binary CT image. The neutral axis of the image, which was assumed to be perpendicular to the loading direction, was found by first calculating the area of bone in each row of the image, which is the sum of ones in a given row multiplied by the pixel dimensions in millimeters.

**Table 1 Subject Information and Anthropometric Data**

| Test ID | Aspect | Subject Number | Gender | Age (years) | Specimen Length (cm) |
|---|---|---|---|---|---|
| 1 | Right | 1 | F | 61 | 13.2 |
| 2 | Left | 2 | M | 89 | 14.5 |
| 3 | Right | 3 | M | 75 | 13.8 |
| 4 | Left | 4 | M | 66 | 15.7 |
| 5 | Left | 5 | M | 84 | 15.7 |
| 6 | Left | 6 | M | 84 | 15.4 |
| 7 | Right | 7 | F | 64 | 16.9 |
| 8 | Right | 8 | M | 67 | 14.9 |
| 9 | Right | 9 | F | 92 | 14.3 |
| 10 | Left | 10 | M | 45 | 16.5 |
| 11 | Left | 1 | F | 61 | 13.2 |
| 12 | Right | 2 | M | 89 | 14.8 |
| 13 | Left | 3 | M | 75 | 13.8 |
| 14 | Right | 4 | M | 66 | 15.2 |
| 15 | Right | 5 | M | 84 | 16.0 |
| 16 | Right | 6 | M | 84 | 15.4 |
| 17 | Left | 7 | F | 64 | 16.7 |
| 18 | Left | 8 | M | 67 | 15.7 |
| 19 | Left | 9 | F | 92 | 14.6 |
| 20 | Right | 10 | M | 45 | 15.9 |

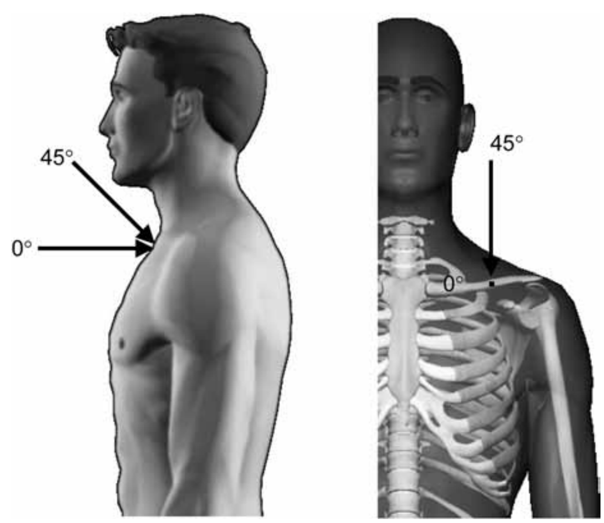


**Figure 1** — Clavicle impact locations, left, and anatomical positions, right.

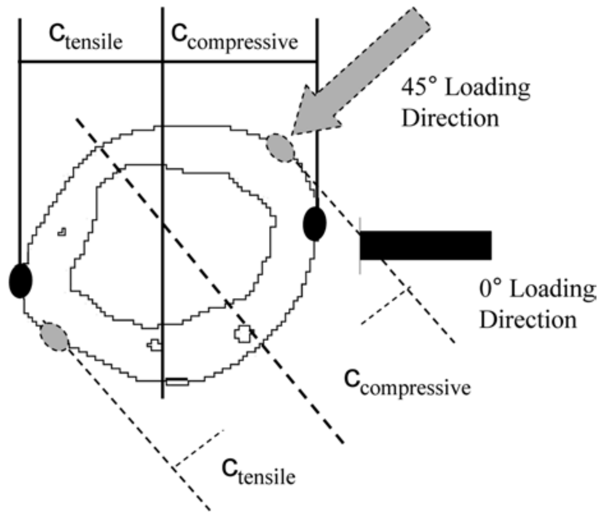


**Figure 2** — Sample image outline and assumptions for processing.

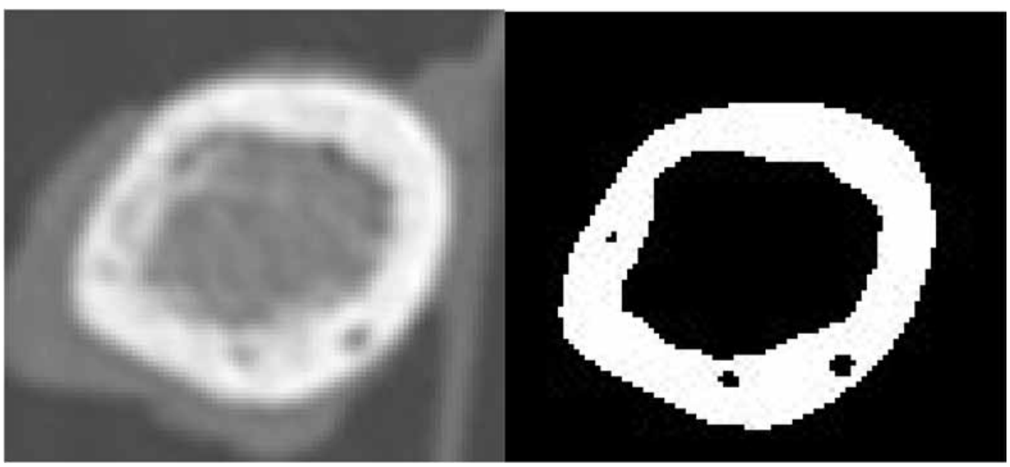

**Figure 3** — Sample clavicle cross section before (left) and after (right) thresholding.

The area of bone in each row of pixels was multiplied by the distance that row was from an assumed neutral axis, and then summed above and below the assumed neutral axis. The neutral axis was determined to be the row in which the sum above the assumed neutral axis was equal to the sum below the assumed neutral axis. The distance to the neutral axis (c) was calculated by multiplying the number of rows between the neutral axis and the tensile side by the millimeters per pixel value. The area moment of inertia (I) about the neutral axis was found by calculating the area of bone in a row multiplied by the square of the distance that row was from the neutral, and then summing that value up for the entire image. The area of cortical bone (A) was calculated as the by sum of ones over the entire image multiplied by the pixel dimensions in millimeters.

The primary component of the test setup was a servo-hydraulic Material Testing System (MTS 810, 13.3 kN; Eden Prairie, MN; Figure 4). The MTS used a hydraulic actuator and impactor head to load the clavicles in a dynamic three-point bending configuration at a displacement rate of 152 mm/s. This corresponded to a strain rate of 0.3 strain/s at the midpoint of the tensile side of the clavicle, which is consistent with the thoracic belt loading test results reported by Duma et al. (2005). Displacement was measured using the MTS internal linear variable displacement transducer (LVDT). A six-axis impactor load cell (Denton 1968, 22,240 N; Rochester Hills, MI) was used to measure loads exerted on the specimen by the impactor. An accelerometer (Endevco 7264B, 2,000 *g*; San Juan Capistrano, CA) was attached to the impactor head to allow for inertial compensation of the impactor load cell, using the mass between the clavicle and active axis of the load cell. The support structure was mounted to a three-axis reaction load cell (Denton 5768, 11,120 N).

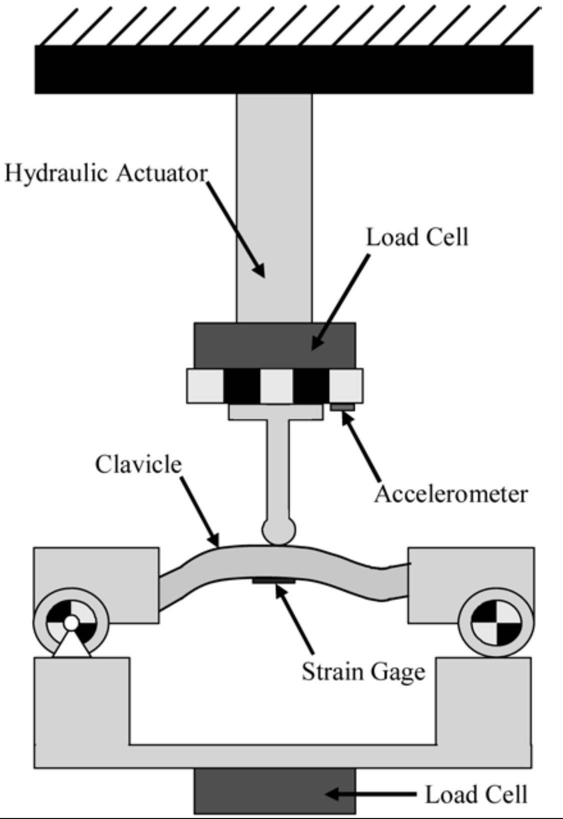


**Figure 4** — Clavicle three-point bending test setup.

To stabilize the clavicle in the three-point test configuration while maintaining loading in the desired plane, the ends were placed in rigid square aluminum potting cups, containing polymer filler (Bondo Corporation, Atlanta, GA). Special care was taken during the potting process to ensure that the clavicles were oriented to produce loading in the desired direction when placed on the three-point bending test setup. The sternoclavicular joint is a saddle type of synovial joint but it functions like a ball-and-socket joint, allowing rotation about all three axes but not translation (Moore and Dalley, 2006; Terry and Chopp, 2000). Therefore, a cylindrical polymer roller was attached to the medial potting cup and pinned to allow for rotation but not translation. The acromioclavicular joint is a plane type of synovial joint, but movements at this joint are similar to those of the sternoclavicular joint (Williams et al., 1989). Specifically, the acromioclavicular joint allows axial rotation, anterior-posterior rotation, and superior-inferior rotation (Branch et al., 1996). In addition, since the sternoclavicular joint is the only articulation between the upper limb and the axial skeleton, the clavicle in its movement carries with it the scapula, which glides on the exterior portion of the thorax. As a result, the lateral end of the clavicle is free to translate in all directions. Therefore, a cylindrical polymer roller was attached to the lateral end of the clavicle but not constrained laterally.

After potting, each clavicle was instrumented with a uniaxial strain gauge (Vishay Measurements Group, CEA-06–062UW-350; Malvern, PA) mounted to the tensile side, which was the side opposite the impactor, at the midpoint of the specimen (Kemper et al., 2007a). For specimens impacted at 0° from the transverse plane, the tensile side corresponded to the posterior portion of the clavicle. For specimens impacted at 45° from the transverse plane, the tensile side corresponded to the inferior-posterior portion of the clavicle. It should be noted that normal saline was sprayed directly on the samples to maintain proper specimen hydration during both specimen preparation and testing (Reilly et al., 1974).

All data channels were recorded at a sampling frequency of 20 kHz (Iotech WBK16; Cleveland, OH). Data from the load cells and accelerometers were filtered to channel filter class (CFC) 180. The moment was calculated as one half the inertially compensated impactor force multiplied by one half of the active span, which corresponds to the initial distance between the centers of the two rollers. The stiffness (K) was determined by performing a linear regression on the initial linear region of the force versus displacement curve ($R^2 > .9$), approximately 10–40% of the peak force. After testing, the distance from the center of the strain gauge to the point of fracture on the tension side was documented. In the case of a comminuted fracture, the measurement was taken from the closest fracture location.

Statistical analyses were performed on the structural response variables by performing a paired two-sample *t* test for means. In addition, statistical analyses were performed on the geometric variables: moment of inertia (I), distance to the neutral axis (c), and cortical bone area (A) by performing a paired two-sample *t* test for the means. Statistical significance was defined as $p \leq 0.05$. The goal of the statistical analysis was to determine if there are any statistical differences in structural properties or geometry with respect to impact direction.

## Results

The clavicle strain time histories in the current study were found to be similar to that of observed shoulder belt loading performed on intact postmortem human subjects. This similarity was determined by comparing the strain time histories obtained in the clavicle three-point bending tests performed in the current study to those achieved in the male and female shoulder belt tests conducted by Duma et al. (2005) (Figure 5).

All clavicle fractures occurred at approximately the midpoint between the supports. The clavicles fractured in the form of a transverse, oblique, or comminuted wedge-shaped fracture, which are typical of three-point loading conditions (Figure 6). The force versus deflection responses were found to be similar for both test

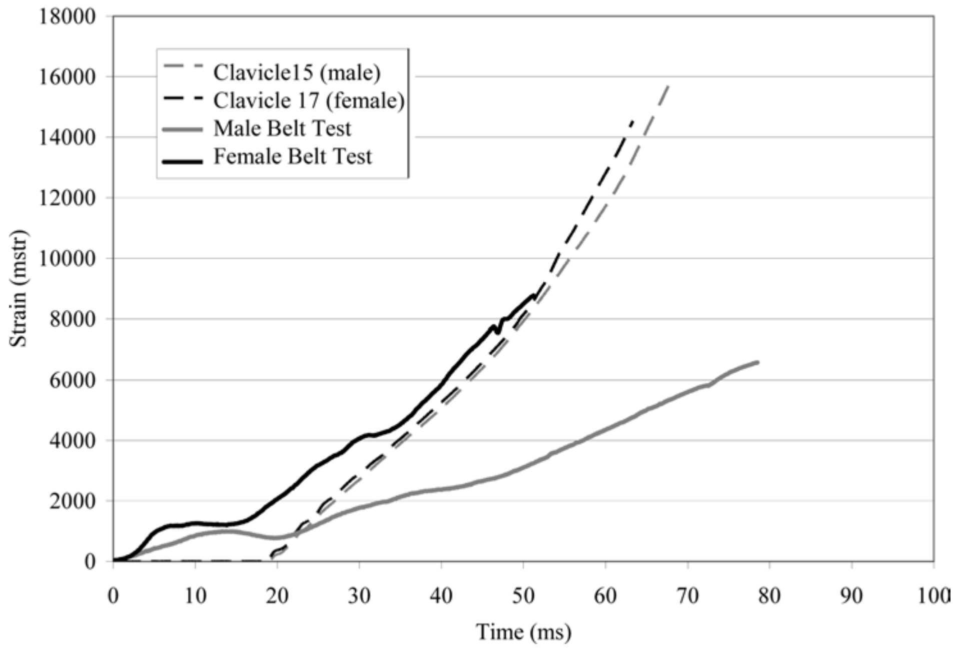


**Figure 5** — Comparison of strain rates between clavicle tests and belt loading tests

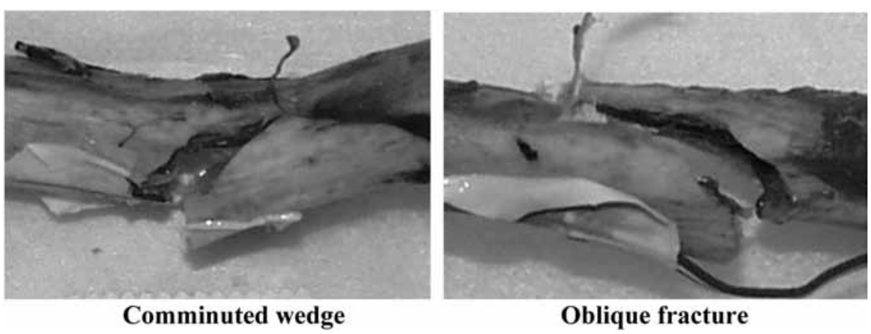


**Figure 6** — Clavicle fracture types resulting from three-point bending.

groups (Figures 7 and 8). The average strain rate for all specimens was 0.3 strain/s.

The peak values were defined as the point at which structural compromising occurred, in other words, the point at which further motion of the impactor produced no incremental increase of applied force (Stein and Granik, 1976). The results indicate that there is no statistically significant difference in peak force ($p$ = .22), peak moment ($p$ = .30), or peak displacement ($p$ = .44) between specimens impacted at 0° versus 45° from the transverse plane (Table 2). However, structural stiffness ($p$ = .01) and peak strain ($p$ < .01) were found to be statistically different between specimens impacted at 0° versus 45° from the transverse plane. The average peak structural values and corresponding standard deviations for both test groups were recorded (Tables 3 and 4).

The results indicate that the area moment of inertia (I; $p$ = .05) and distance to the neutral axis (c; $p$ = .01) were significantly different between specimens impacted at 0° versus 45° from the transverse plane (Table 5). However, neither the ratio of the distance to the neutral axis to the moment of inertia ($p$ = .60) nor the cortical bone area (A; $p$ = .76) were found to be significantly different between the two test groups. The average geometric values and corresponding standard deviations for both test groups were recorded (Tables 6 and 7).

## Discussion

This study investigated the effect of loading direction on the structural response of the human clavicle subjected to dynamic three-point bending. This was accomplished by performing 20 matched tests on whole human clavicles in two loading directions: 0° and 45° from the transverse plane. The controlled data set and specimen

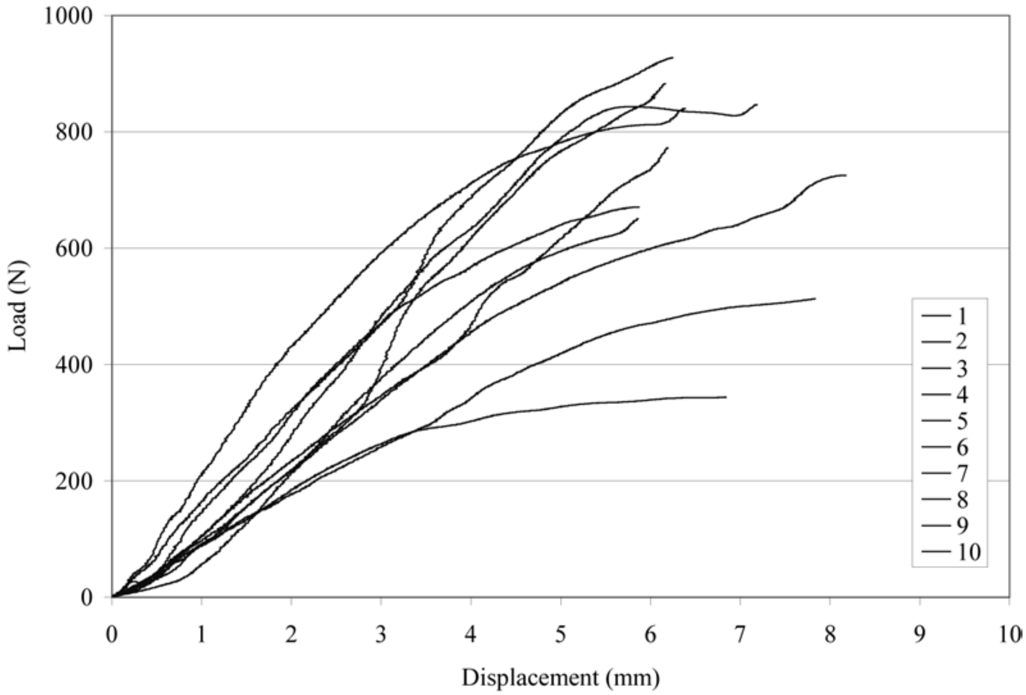


**Figure 7** — Impactor force vs. displacement for all 0° impact tests.

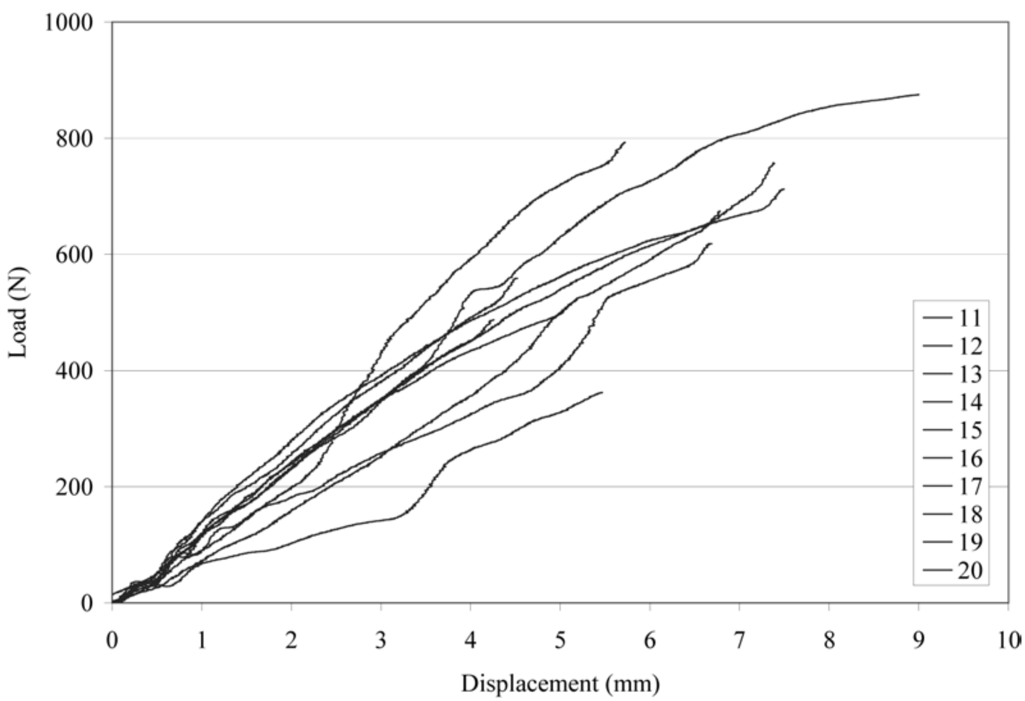


**Figure 8** — Impactor force vs. displacement for all 45° impact tests.

preparation techniques allowed for direct comparison between the two test conditions.

Although the results of the current study where found to be consistent with those reported by previous researchers, an accurate comparison between studies is confounded by differences in end conditions, loading rates, and specimen orientation (Table 8). For example, the average peak force in the current study was consistent with Bolte et al. (2000), but the loading rate in the current study was significantly larger. It is well known

**Table 2 Statistical Analysis of Structural Response Variables for 0° Versus 45° Tests**

| Compression Group | Two-Tail *p*-values | | | | |
|---|---|---|---|---|---|
| | Peak Force | Peak Moment | Peak Displacement | Peak Strain | Stiffness |
| 0° vs. 45° | 0.22 | 0.30 | 0.44 | <0.01* | 0.01 |

*Note.* *Indicates that analysis of peak strain should be regarded with caution.

**Table 3 Peak Structural Response Values for 0° Tests**

| Test ID | Active Span (m) | Peak Impactor Force (N) | Peak Impactor Moment (N·m) | Peak Deflection (mm) | Stiffness (N/mm) | | Peak Strain (mstr) | Gage to Fracture (mm) |
|---|---|---|---|---|---|---|---|---|
| 1 | 0.137 | 840 | 28.8 | 6.4 | 227.1 | * | 19870 | 6.0 |
| 2 | 0.148 | 651 | 24.1 | 5.9 | 163.9 | | 19156 | 0.0 |
| 3 | 0.142 | 670 | 23.8 | 5.9 | 165.7 | * | 21096 | 4.0 |
| 4 | 0.159 | 883 | 35.1 | 6.2 | 128.3 | * | 18059 | 7.0 |
| 5 | 0.162 | 725 | 29.4 | 8.2 | 122.2 | * | 25443 | 6.0 |
| 6 | 0.159 | 513 | 20.4 | 7.8 | 83.3 | * | 18973 | 6.0 |
| 7 | 0.169 | 772 | 32.6 | 6.2 | 123.5 | | 14282 | 16.0 |
| 8 | 0.154 | 847 | 32.6 | 7.2 | 196.8 | | 17087 | 8.0 |
| 9 | 0.146 | 344 | 12.6 | 6.8 | 93.2 | * | 22919 | 16.0 |
| 10 | 0.163 | 928 | 37.8 | 6.2 | 159.0 | * | 20498 | 2.0 |
| Average: | | 717 | 27.7 | 6.7 | 146.3 | | 19738 | 7.1 |
| Standard Deviation: | | 181 | 7.6 | 0.8 | 44.9 | | 3085 | 5.3 |

*Note.* *Indicates that strain gauge failed before peak load.

**Table 4 Peak Structural Response Values for 45° Tests**

| Test ID | Active Span (m) | Peak Impactor Force (N) | Peak Impactor Moment (N·m) | Peak Deflection (mm) | Stiffness (N/mm) | | Peak Strain (mstr) | Gage to Fracture (mm) |
|---|---|---|---|---|---|---|---|---|
| 11 | 0.138 | 517 | 17.8 | 5.1 | 129.2 | | 18065 | 6.0 |
| 12 | 0.150 | 793 | 29.7 | 5.7 | 105.5 | | 17223 | 10.0 |
| 13 | 0.140 | 712 | 24.9 | 7.5 | 121.2 | * | 17556 | 6.0 |
| 14 | 0.156 | 487 | 19.0 | 4.3 | 121.5 | | 10163 | 17.0 |
| 15 | 0.161 | 756 | 30.4 | 7.4 | 144.8 | | 15898 | 7.0 |
| 16 | 0.161 | 674 | 27.1 | 6.8 | 95.7 | | 16664 | 23.0 |
| 17 | 0.170 | 618 | 26.3 | 6.7 | 72.3 | | 14536 | 17.0 |
| 18 | 0.164 | 559 | 22.9 | 4.5 | 131.5 | | 9463 | 2.0 |
| 19 | 0.155 | 362 | 14.0 | 5.5 | 39.9 | | 14184 | 26.0 |
| 20 | 0.175 | 875 | 38.3 | 9.0 | 116.0 | * | 19230 | 1.0 |
| Average: | | 635 | 25.1 | 6.2 | 107.8 | | 15298 | 11.5 |
| Standard Deviation: | | 157 | 7.0 | 1.5 | 31.4 | | 3273 | 8.7 |

*Note*. *Indicates that strain gauge failed before peak load.

that bone is a viscoelastic material, and therefore is affected by the rate of loading (McElhaney and Byars, 1965; Crowninshield and Pope, 1974; Wright and Hayes, 1976; Wood, 1971; Carter & Haynes, 1976; and Stein & Granik, 1976). Assuming that all other test conditions were similar, one would expect the peak force to increase with loading rate. However, the end conditions were different between the two studies. Therefore, the

**Table 5 Statistical Analysis of Cross-Sectional Geometry for 0° Versus 45° Specimens**

| | Two-Tail *p*-values | | | |
|---|---|---|---|---|
| **Compression Group** | **I** | **c** | **c/I** | **A** |
| 0° vs. 45° | 0.05 | 0.01 | 0.60 | 0.76 |

**Table 6 Cross-Sectional Geometry Values for Specimen Impacted at 0°**

| Test ID | Impact Angle | I ($mm^4$) | c (mm) | c/I ($mm^{-3}$) | A ($mm^2$) |
|---|---|---|---|---|---|
| 1 | 0 | 542.8 | 5.6 | 0.010 | 65.4 |
| 2 | 0 | 737.8 | 6.3 | 0.009 | 48.1 |
| 3 | 0 | 439.9 | 5.5 | 0.013 | 49.8 |
| 4 | 0 | 881.8 | 6.6 | 0.007 | 57.6 |
| 5 | 0 | 546.6 | 5.6 | 0.010 | 49.3 |
| 6 | 0 | 292.1 | 5.3 | 0.018 | 26.5 |
| 7 | 0 | 940.5 | 6.6 | 0.007 | 72.9 |
| 8 | 0 | 832.8 | 5.9 | 0.007 | 46.0 |
| 9 | 0 | 373.1 | 6.2 | 0.017 | 28.5 |
| 10 | 0 | 896.3 | 6.0 | 0.007 | 68.5 |
| Average: | | 648.4 | 6.0 | 0.010 | 51.3 |
| Standard Deviation: | | 238.3 | 0.5 | 0.004 | 15.6 |

**Table 7 Cross-Sectional Geometry Values for Specimen Impacted at 45°**

| Test ID | Impact Angle | I ($mm^4$) | c (mm) | c/I ($mm^{-3}$) | A ($mm^2$) |
|---|---|---|---|---|---|
| 11 | 45 | 326.9 | 4.1 | 0.010 | 64.4 |
| 12 | 45 | 544.3 | 5.3 | 0.013 | 40.4 |
| 13 | 45 | 347.4 | 4.3 | 0.010 | 54.2 |
| 14 | 45 | 548.3 | 5.8 | 0.012 | 48.4 |
| 15 | 45 | 741.4 | 5.4 | 0.011 | 65.6 |
| 16 | 45 | 352.0 | 4.5 | 0.007 | 40.1 |
| 17 | 45 | 503.6 | 4.4 | 0.013 | 66.6 |
| 18 | 45 | 674.1 | 5.4 | 0.009 | 46.9 |
| 19 | 45 | 284.9 | 5.5 | 0.008 | 27.2 |
| 20 | 45 | 832.1 | 6.5 | 0.019 | 67.3 |
| Average: | | 515.5 | 5.1 | 0.011 | 52.1 |
| Standard Deviation: | | 189.2 | 0.8 | 0.004 | 13.8 |

**Table 8 Comparison of Current Study With Previous Clavicle Three-Point Bending Literature**

| | | Impact Direction | Loading Rate | Average Peak Force | Average Stiffness |
|---|---|---|---|---|---|
| **Author** | **End Conditions** | **Degrees from Transverse Plane** | **mm/s** | **N** | **N/mm** |
| Bolte et al. (2000) | Evenly Supported | N/R | 0.50 | 667 (± 277) | N/R |
| Proubasta et al. (2002) | Fixed-Fixed | 0 ° | 0.05 | 486 (± 179) | 94.8 (± 4.9) |
| Current Study | Simply Supported | 0 ° | 152.00 | 717 (± 181) | 146.3 (± 44.9) |
| | Simply Supported | 45 ° | 152.00 | 635 (± 157) | 107.8 (± 31.4) |

*Note.* N/R = not reported.

two studies cannot be directly compared owing to the possible variations in applied shear and/or axial torsion, both of which would effectively decrease the measured peak force. Regardless, the findings of previous authors are presented for completeness.

With respect to loading direction, the results of the current study showed there were no statistically significant differences in peak force ($p$ = .22), peak moment ($p$ = .30), or peak displacement ($p$ = .44) between specimens impacted at 0° versus 45° from the transverse plane. In contrast, the structural stiffness ($p$ = .01) and peak strain ($p$ < .01) were found to be significantly different between specimens impacted at 0° versus 45° from the transverse plane. However, the location of the fracture on the tension side of the specimens rarely corresponded to the location of the strain gauge because of the complex fracture types. In addition, the strain gauge occasionally failed before peak load. For these two reasons, the peak strain reported in this article is less than the actual peak strain. The degree to which the reported peak strain varies from the actual is related to the distance from the point of fracture to the center of the strain gauge, as a result of the strain distribution on the tensile side of the specimen, and the point at which the gauge failed. Therefore, peak strain is not an appropriate comparison variable and should be regarded with caution.

Variations in the structural response of whole bone sections can be a result of changes in the bone geometry, bone material properties, or both. Due to the controlled matched data set, the differences in the structural response with respect to loading direction can be attributed to geometric differences and not material differences. The analysis of the clavicle cross-sectional geometry showed that the area moment of inertia ($p$ = .05) and distance to the neutral axis ($p$ = .01) were significantly different between specimens impacted at 0° versus 45° from the transverse plane. However, the ratio of the distance to the neutral axis to the moment of inertia ($p$ = .60) and cross-sectional area ($p$ = .76) was not found to be significant with respect to loading direction. Therefore, differences in the structural stiffness were most likely a result from the overall geometry of the clavicle and not differences in the cross-sectional geometry. To explain, the clavicle has a complex geometry that can best be described as an S-shape. If there is significant curvature, then the neutral axis will no longer coincide with the neutral axis and the stress distribution across the section will become more hyperbolic (Norton, 2000). In the case that the loading is applied perpendicular to the radius of curvature, the neutral axis will shift toward the center of curvature and more of the specimen will be placed in compression. Previous researchers have shown that elastic modulus of bone is lower in tension than compression (Kemper et al., 2007b; Burstein et al., 1976; Reilly & Burstein, 1975; Evans & Bang, 1967). Therefore, loading a curved bone in different directions could result in changes to the overall structural stiffness.

The results of the current study suggest that optimizing seatbelt designs to load the midpoint of the clavicle at 0° versus 45° from the transverse plane, or vise versa, during frontal belt loading would have no effect on whether a fracture will occur. However, there are other variables that should be evaluated to determine if it is possible to optimize seatbelt designs to minimize the risk of clavicle fractures. Specifically, the location, medial or lateral of the midpoint, and distribution of the applied force could significantly affect the failure strength of the clavicle. Therefore, future studies should be conducted to investigate the effects of these variables to further the understanding of the biomechanical response of the clavicle in three-point bending. The results of such studies would provide researchers and safety engineers with valuable data that could lead to an optimized of seatbelt restraint system to minimize the risk of clavicle fractures in frontal belt loading.

In conclusion, the results of the current study showed that the only significant difference between 0° and 45° impacts was the stiffness. Specifically, clavicles impacted 45° from the transverse plane were less stiff than the clavicles impacted 0° from the transverse plane. Due to the controlled matched data set, the difference in the structural response with respect to loading direction can be attributed to the complex geometry of the clavicle and not material differences. Finally, addition testing should be conducted to investigate the effects of the location and distribution of the applied load to more fully understand the biomechanical response of the clavicle in three-point bending.

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