Impact Response and Simulation of Damaged Ulna With Internal Fixation

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The objectives of this work were to explore a methodology that combines static and dynamic finite element (FE) analysis, linear elastic fracture mechanics (LEFM) and experimental methods to investigate a worst-case scenario in which a previously damaged bone plate system is subjected to an impact load. Cadaver ulnas with and without midshaft dynamic compression plates are subjected to a static three-point bend test and loaded such that subcritical crack growth occurs as predicted by a hybrid method that couples LEFM and static FE. The plated and unplated bones are then unloaded and subsequently subjected to a midshaft transverse impact test. A dynamic strain-based FE model is also developed to model the midshaft transverse impact test. The average value of the impact energy required for failure was observed to be 10.53% greater for the plated set. There appears to be a trade-off between impact damage and impact resistance when ulnas are supported by fixation devices. Predictions from the dynamic FE model are shown to corroborate inferences from the experimental approach.

Keywords: finite element, bone, fracture, crack, failure

Dynamic compression plate (DCP) fixation systems have been successfully used in the repair of long bone trauma for the last several decades. In general, these systems have proven to be reliable; however, they are not without their drawbacks. The reported mechanical failures include screw pull-out (Hyldahl et al., 1991), corrosion of plates and/or screws, fatigue cracks (Busam et al., 2006), fatigue cracks exacerbated by corrosion (Zand et al., 1983), and cracks initiated by low impact loads and exacerbated by fatigue and/or corrosion (Thomas et al., 1988). Only a few site-specific cases of refractures of long bones with retained plates have been reported in the literature (McLean et al., 2007); however, between 1990 and 2005, the frequency of orthopedic implant reoperations per 10,000 fall admissions (on level surfaces) of hip, knee, spine, or fracture fixation implants increased by 35% from 79.0 (95% CI: 65.2–92.8) to 106.7 (95% CI: 97.9–115.4). The frequency of orthopedic implant reoperations per 10,000 fall admissions (from stairs) increased 36% from 48.9 (95% CI: 21.5–76.3) to 66.7 (95% CI: 52.3–81.1). In 2005, 95% (level surfaces) and 93% (stairs) of these reoperations involved fracture fixation implants (Ong et al., 2009). The impact response of cortical bone has been examined by several authors (Wakao et al., 2009, Thambyah et al., 2008, Troy & Grabiner, 2007; Currey, 1979, Reilly & Curry, 2000). In addition, the impact strength of materials has been considered in their selection for internal fixation (Shikinami & Okuno, 1999, Tams et al., 1995, Grijpma et al., 1992). There is no consensus within the orthopedic community whether metalwork in asymptomatic skeletally mature adults should be removed after fracture healing. Although the literature in this area is relatively scant, the most recent recommendations trend toward plate retention, citing the risks of peri-implant refracture, metal sensitivity and carcinogens as low (Busam et al., 2006). In another recent survey, 92% of orthopedic surgeons indicated that they do not routinely remove metalwork in asymptomatic skeletally mature patients (Jamil et al., 2008). As the percentage of active older adults continues to rise and internal fixation methods become more sophisticated and routine, hardware retention will also most likely increase. Given that the evidence guiding hardware removal decisions and risks of retention is quite limited, especially for specific cases (e.g., a both-bone forearm fracture repair), it would be prudent to explore the impact response of bone and plate when the implants are retained after fracture healing. Reilly and Curry (2000) examined the effects of damage and microcracking on impact strength of bone. These authors performed experimental tests on bovine bone and found that tensile damage in bones loaded in tension did not reduce the bone’s energy-absorbing ability until the occurrence of a modulus reduction of over 20%. However, compression damage for bones loaded in tension severely reduced the bone’s energy absorption capability by an average of about 40%. There is a strong
potential that inferences from these types of studies could assist orthopedic surgeons in their initial recommendation for hardware removal or retention, with the goal of reducing surgical complexity during peri-implant fracture reduction, medical costs and minimization of patient days away from work. In the application of fixation implants, knowledge of their effect on the mechanical limits of the bone (i.e., worst-case scenario) would be an asset to the orthopedic surgeon. Results may be particularly applicable to patients with retained fixation implants whose professional vocation involves constant risk of physical trauma, such as professional athletes (football players, mixed martial artists, boxers) or workers who interact with heavy impact machinery (construction, oil platform workers, etc.) on a daily basis.

If mechanical limits are redefined to include worst-case scenarios, the question of the effect of the implant on the mechanical response of the bone due to impact loads now becomes an important one. Since microcracks develop more easily in older bone, we should also explore the effect of microcracking on the impact strength of the bone/plate system. In this work, the authors explore the scenario where the bone and fixation system have been impacted at a site that was previously subjected to static load/unload. While the likelihood of bones being subjected to a three-point bend static load and subsequent impact at the same site is remote, this is a worst-case scenario, the examination and understanding of which these authors believe will enable engineers to fully define the design envelope for improved fixation system performance. The authors also explore the capability of a method that combines fracture mechanics predictions with experimental testing and computational methods to investigate the impact response of previously damaged ulnas with and without internal fixation. The following sections present further details on the methods used: linear elastic fracture mechanics (LEFM), finite element (FE) analysis and experimental procedure, followed by the results and discussion.

**Methods**

**Overview**

The methods are summarized in chronological order in Figure 1. Static FE models of the ulna bone with and without fixation subjected to load conditions which simulate a three-point bend test were first developed. Stress results from the static FE model were combined with LEFM to predict the crack growth instability load to be used in a static three-point bend test, the purpose of which was to ensure that some degree of damage occurs in the specimens. Specimens were then subjected to the static three-point bend test and loaded up to the theoretical (LEFM-derived) crack growth instability point. Dynamic FE models were then developed to investigate the impact response.

![Figure 1 — Chronological description and relationships between methods.](image-url)
Finite Element Analysis: Static Three-Point Bend Tests

The FE analysis was performed using CosmosWorks (Dassault System, 2008) for the static case. The ulna was modeled as a hollow cylindrical shell representing the cortical bone with an inner hollow cylindrical shell representing the cancellous bone. The geometry of the bone and internal fixation device used in the model are provided in Table 1.

The model was discretized using a fine mesh with 14,600 parabolic triangular shell elements; a fine mesh was necessary to avoid element warpage and simultaneously incorporate element failure criteria (strain based) in the impact model. Two simulations with higher mesh densities were performed; however, nodal stress results were unaltered, indicating that mesh convergence had already been achieved with 14,600 elements. The cylinder ends were fixed in the respective degrees of freedom used in the experimental set-up for the static test. The loading was applied transversely at the ulna mid-diaphysis. A linear orthotropic material model was used for both the cortical and cancellous bone. The material property values used (Krone & Schuster, 2006) are shown in Table 2. The orthotropic model assumes that both the cortical and cancellous bones are stronger along the longitudinal direction than in the transverse directions.

Linear Elastic Fracture Mechanics

Several authors (Akkus et al., 1999) report that crack initiation along the crack plane in cortical bone is precipitated by damage at the ultra-structural level where interstitial microcracks simultaneously initiate and accumulate at neighboring sections. In order determine a reasonable load value at which the bone would have sustained damage at the ultra-structural level, LEFM methods were employed to determine the crack growth instability load during quasi-static loading. A conservative approach was adopted that utilizes LEFM applications for circumferential through the wall cracks in hollow cylindrical pipes (Takahashi, 2002; Lacire et al., 1999). While Ritchie et al. (2008) demonstrated the application of these LEFM methods for rat and femur bones, the method itself is dependent on geometry only and was derived for cylindrical pipes in which $1.5 < R_m/t < 80.5$ and $0 < \Theta / \pi < 0.61$, where $t$ is the cortical thickness, $R_m$ is the mean radius of the bone (to the middle of the cortex) and $\Theta$ is the half-crack angle. Human bone geometries are also well within these ranges. It was shown to suitably model small animal bones for fracture toughness determination mainly due to the fact that their physiological size disallows crack growth monitoring needed for more qualitatively descriptive crack growth resistance methods (e.g., $R$-curve). However, as this approach ignores toughening mechanisms acting in the crack wake (Ritchie, 1988, Ritchie et al., 2008, Nalla et al., 2003) as well as additional plasticity effects, application of this method has been shown to under-predict fracture toughness. The average age of the human bones tested in this research is relatively high (83.7 years), so it is likely that the bones already contain preexisting microcracks. A low fracture toughness value therefore allows for stable crack growth of any preexisting microcracks and any decrease in fracture toughness due to bone age. The latter is predicted to be 4.1% per decade after age 35 (Zioupos & Currey, 1998).

The fracture toughness, $K_c$, of cortical bone reported in the literature varies between 2 and 6 MPa$\sqrt{\text{m}}$ (Zioupos & Currey, 1998, Nalla et al., 2005, Yang et al., 2006). A conservative value of 2 MPa$\sqrt{\text{m}}$ was used in Equation (1) to calculate the applied bending stress, $\sigma_c$, necessary for crack growth instability. In Equation (1), $F_b$ is a geometry factor that depends on the ratio of the thickness, $t$, to the mean radius $R_m$. The values used in the analysis were $R_m = 9 \text{ mm}$, $\Theta = \pi/6 \text{ rad}$ and $t = 2 \text{ mm}$. The values of $F_b$ and

Table 2  Material properties used in ulna FE model

<table>
<thead>
<tr>
<th>Property</th>
<th>Material</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Density, $p$ (kg/m$^3$)</td>
<td>Cortical Bone</td>
<td>1900</td>
</tr>
<tr>
<td></td>
<td>Cancellous Bone</td>
<td>400</td>
</tr>
<tr>
<td>$E_1$ (GPa)</td>
<td>Cortical Bone</td>
<td>16</td>
</tr>
<tr>
<td></td>
<td>Cancellous Bone</td>
<td>80</td>
</tr>
<tr>
<td>$E_2$ (GPa)</td>
<td>Cortical Bone</td>
<td>6.8</td>
</tr>
<tr>
<td></td>
<td>Cancellous Bone</td>
<td>60</td>
</tr>
<tr>
<td>$E_3$ (GPa)</td>
<td>Cortical Bone</td>
<td>6.8</td>
</tr>
<tr>
<td></td>
<td>Cancellous Bone</td>
<td>60</td>
</tr>
<tr>
<td>$G_{12}$ (GPa)</td>
<td>Cortical Bone</td>
<td>3.2</td>
</tr>
<tr>
<td></td>
<td>Cancellous Bone</td>
<td>26</td>
</tr>
<tr>
<td>$G_{13}$ (GPa)</td>
<td>Cortical Bone</td>
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<td></td>
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<td>37</td>
</tr>
<tr>
<td>$G_{23}$ (GPa)</td>
<td>Cortical Bone</td>
<td>3.3</td>
</tr>
<tr>
<td></td>
<td>Cancellous Bone</td>
<td>52</td>
</tr>
<tr>
<td>$V_{12}$</td>
<td>Cortical Bone</td>
<td>0.30</td>
</tr>
<tr>
<td></td>
<td>Cancellous Bone</td>
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</tr>
<tr>
<td>$V_{13}$</td>
<td>Cortical Bone</td>
<td>0.30</td>
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<tr>
<td></td>
<td>Cancellous Bone</td>
<td>30</td>
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<tr>
<td>$V_{23}$</td>
<td>Cortical Bone</td>
<td>0.30</td>
</tr>
<tr>
<td></td>
<td>Cancellous Bone</td>
<td>30</td>
</tr>
<tr>
<td>Failure strain $\varepsilon_f$</td>
<td></td>
<td>0.01</td>
</tr>
</tbody>
</table>

Table 1  Geometry of bone and fixation system (mm)

<table>
<thead>
<tr>
<th>Type</th>
<th>Length</th>
<th>Outer Diameter</th>
<th>Thickness</th>
<th>Width</th>
</tr>
</thead>
<tbody>
<tr>
<td>Plate</td>
<td>160</td>
<td>N/A</td>
<td>3.5</td>
<td>10</td>
</tr>
<tr>
<td>Cortical Bone</td>
<td>230</td>
<td>35</td>
<td>3</td>
<td>N/A</td>
</tr>
<tr>
<td>Cancellous Bone</td>
<td>230</td>
<td>22</td>
<td>6</td>
<td>N/A</td>
</tr>
</tbody>
</table>
\( R_m \) were estimated based on the typical ulna cortical thicknesses found in the literature (Hsu et al., 1993) and the measured outer radius. The half crack angle was conservatively estimated based on carbon steel circumferential cracks (higher fracture toughness) of similar geometries (Kashima, 2002). The latter estimation implies that the critical load derived would allow stable crack growth up to this crack geometry.

\[
\sigma_b = \frac{K_c}{F_b \sqrt{\pi R_m}} \tag{1}
\]

where \( F_b \) is given by

\[
F_b = \left(1 + \frac{t}{R_m}\right) \left[ A_b + B_b \left(\frac{\Theta}{\pi}\right) + C_b \left(\frac{\Theta}{\pi}\right)^2 + D_b \left(\frac{\Theta}{\pi}\right)^3 + E_b \left(\frac{\Theta}{\pi}\right)^4 \right] \tag{2}
\]

and

\[
A_b = 0.65133 - 0.5774s - 0.3427\xi^2 - 0.0681e^3
\]

\[
B_b = 1.879 + 4.795s + 2.343\xi^2 - 0.6197e^3
\]

\[
C_b = -9.779 - 38.14s - 6.611\xi^2 + 3.972e^3
\]

\[
D_b = 34.56 + 129.9s + 50.55\xi^2 + 3.374e^3
\]

\[
E_b = -30.82 - 147.6s - 78.38\xi^2 - 15.54e^3
\]

\[
\xi = \log\left(\frac{t}{R_m}\right)
\]

The uniqueness of the \( t/R_m \) ratio for each bone resulted in different loads required for crack growth instability. The presence of the plate and screws will serve to decrease tensile stresses along the cross-section (trauma location) when the bone is subjected to pure bending (plate on tensile side), which may increase the critical load value needed to achieve \( K_c \). However, the stress gradient will have increased due to the shift of the neutral axis toward the plates and higher compressive stresses closer to the plate. The steeper stress gradients created in addition to the stress concentrations due to the holes may decrease \( K_c \). The difference in \( K_c \) due to the plate was found by inserting a crack in the plated and unplated FE models. Estimates of \( K_c \), the stress intensity factor, at different distances from the crack tip were then obtained based on the stress values along the crack axis but ahead of the crack tip and plotted as a function of the distance, \( r \), from the crack tip. A value for \( K_c \) at the crack tip was found by extrapolating a fitted curve to \( r = 0 \). Using this approach (Fischer-Cripps, 2000), the plated bone had a reduced \( K_c \) by approximately 80% compared with the unplated. Therefore, a similar reduction was applied to \( K_c \) to determine the bending stress at which crack growth instability occurred for the plated bone. The corresponding load was then extrapolated using composite beam theory. These relationships are shown in Equations (3) and (4). For symmetric beams made of \( n \) materials, the bending stress, \( \sigma_j \), in the \( j \)th material is given by

\[
\sigma_j = -\frac{E_j P}{\sum_{i=1}^{n} 4E_i I_i} \tag{3}
\]

where the location of the neutral axis, \( \hat{y} \), is given by

\[
\hat{y} = \frac{\sum_{i=1}^{n} E_i A_i}{\sum_{i=1}^{n} E_i A_i} \tag{4}
\]

and \( P \) represents the transverse point load at the center of the beam, \( \ell \) is the beam length, and \( t \) is the distance from the neutral axis to the point of concern. The terms \( E_i, I_i, A_i, \) and \( \ell_i \) are, respectively, the elastic modulus, moment of inertia with respect to the neutral axis, cross-sectional area and location of the centroidal axis of the \( j \)th beam. Since the added plate geometry increases both the effective \( t \) and the effective \( R_m \) by the same factor, the ratio \( t/R_m \) for the cylinder with attached plates did not change significantly (< 1%). Composite beam theory ignores relative movement between the plate and bone; additionally, it is based on the Euler–Bernoulli beam model, which ignores Poisson’s ratio and assumes that the material is linear elastic, homogenous and isotropic. While bone is anisotropic and inhomogenous, several authors have found composite beam theory applicable in successfully predicting macro response of bone plate systems (Carter et al., 1984; Cordely et al., 2000; Ramakrishna et al., 2004). As this work is exploratory in nature, we sought a correlation among various methods chosen for their simplicity and ease of implementation to minimize computational cost. If the simple theoretical and computational models coupled with experiments provided results consistent with each other as well as the literature, then the effect of the systematic removal of assumptions may be studied in future work.

**Experimental Procedure**

Six fresh frozen adult cadaver ulnas with age ranges between 77 and 90 years old were mechanically tested. The bones were from Caucasian males with no history of severe trauma to their ulnas or severe disease. Plates were attached to three bones (designated as set plated) via cortical screws to simulate a midshaft compound fracture repair system after full healing. The bone designations and characteristics are shown in Table 3. Two of the plates were dynamic compression plates (DCPs) and one, \( P_2 \), was a limited-contact dynamic compression plate (LC-DCP). The remaining three bones (designated as set unplated) did not have an attached fixation system. The outer radius at the mid-shaft of each bone was measured using Vernier calipers. The average bone radius, \( R_m \), used in the LEFM model was estimated from these measurements.

The specimens were prepared by soaking them in air-tight containers with Hanks’s balanced salt solution (HBSS; 1×) solution without phenol red and a 10 mg/mL concentration of gentamicin for a minimum time period of 40 hr at 4 °C, before testing procedures were
Coates et al. performed. The pH was maintained at 7.4 while in solution. During testing procedures, the specimens were constantly irrigated with HBSS to minimize bone desiccation. Testing procedures were conducted at a temperature of 25 °C.

Epoxy end caps were manufactured so that the bones could be held in place during loading and end rotations in the horizontal plane and the vertical plane perpendicular to bone axis would be minimized during testing. The epoxy end caps were sanded along their curved faces to form a flat surface at their bottoms so that the bone would have a stable horizontal position during testing procedures. Epoxy end caps were approximately 6 cm long, with 3 cm covering the longitudinal ends of the bone, and 5 cm in diameter.

**Static Three-Point Bend Tests.** The static three-point bend test was done in accordance with ASTM D790: Standard Test Methods for Flexural Properties of Unreinforced and Reinforced Plastics and Electrical Insulating Materials. An ulna in the three-point bend static test configuration is shown in Figure 2. Standard D790 calls for each sample to be tested based on independent dimensions. Because of this, a crosshead displacement rate had to be calculated based on the depth and length of each sample. The crosshead rates as well as the diameter and length of each sample are also listed in Table 3. The unplated bones were loaded to 540 N (±25 N) and unloaded, plated bones were loaded to 450 N (±30 N) and unloaded. These load values were predicted by the LEFM model for the onset of crack growth instability based on FE values of the stress intensity reduction at the crack tip due to the plates.

**Dynamic Impact Tests.** The dynamic drop weight impact tests were performed in accordance with ASTM

<table>
<thead>
<tr>
<th>Bone Label</th>
<th>Age</th>
<th>Length (mm)</th>
<th>Plate/Type</th>
<th>Plate Length (mm)</th>
<th>Diameter (mm)</th>
<th>Displacement Rate (mm/s)</th>
</tr>
</thead>
<tbody>
<tr>
<td>U₁</td>
<td>80</td>
<td>216</td>
<td>N/A</td>
<td>N/A</td>
<td>17.8</td>
<td>0.073</td>
</tr>
<tr>
<td>U₂</td>
<td>77</td>
<td>286</td>
<td>N/A</td>
<td>N/A</td>
<td>18.4</td>
<td>0.123</td>
</tr>
<tr>
<td>U₃</td>
<td>86</td>
<td>238</td>
<td>N/A</td>
<td>N/A</td>
<td>17.4</td>
<td>0.090</td>
</tr>
<tr>
<td>P₁</td>
<td>82</td>
<td>219</td>
<td>Yes/LC-DCP</td>
<td>85</td>
<td>21.4</td>
<td>0.062</td>
</tr>
<tr>
<td>P₂</td>
<td>90</td>
<td>203</td>
<td>Yes/DCP</td>
<td>115</td>
<td>23.0</td>
<td>0.050</td>
</tr>
<tr>
<td>P₃</td>
<td>87</td>
<td>245</td>
<td>Yes/DCP</td>
<td>125</td>
<td>20.9</td>
<td>0.079</td>
</tr>
</tbody>
</table>

**Figure 2** — Ulna in static three-point bend test.
Finite Element Analysis: Dynamic Impact Tests

The FE analysis was performed using LSDYNA (Livermore Software Technology Corporation, 2008), for the dynamic case. The ulna and the internal fixation devices were modeled similar to that used in the static FE analysis with the ends fixed as per the impact test. The impact load was simulated as a solid, rigid piece of steel (7870 kg/m³ density and 205 GPa elastic modulus) with an impact velocity of 9.5 m/s for the unplated set and 10 m/s for the plated set. The impact edge was circular with a diameter of 5 mm. The dynamic study used approximately 8,400 solid tetrahedral elements. Note that shell elements would have been more efficient in terms of computational cost; however, the isotropic plastic failure criteria (based on failure strain) that the authors deemed most applicable was not available for shell elements. Nalla et al. (2005) provided strong evidence that, in terms of intrinsic damage, the onset of fracture in human cortical bone is consistent with a strain-based criterion. The failure criterion in this work computes the accumulation of strain in the elements, then compares values computed with the predefined failure strain and deletes the element (Kleiven 2006; Astier et al., 2008) if the failure strain is exceeded. The dynamic FE analysis used the isotropic-plastic with failure material model. In this material model, the von Mises yield condition is given by Equation (5):

$$\varphi = \frac{1}{2} \sigma_y \mathbf{S} - \frac{\sigma_y^2}{3}$$

where the first term represents the stress invariant in terms of the deviatoric stresses and \( \sigma_y \), the yield stress, is defined in Equation (6) as a function of the effective plastic strain, \( \dot{\epsilon}_p^\text{eff} \), and plastic hardening modulus, \( E_p \), and \( \sigma_0 \), the initial yield.

$$\sigma_y = \sigma_0 + E_p \dot{\epsilon}_p^\text{eff}$$

The effective plastic strain is defined in Equation (7) as

$$\dot{\epsilon}_p^\text{eff} = \int_0^t \frac{2}{3} d\epsilon_p^\text{eff} d\epsilon_p^\text{eff}$$

and the plastic hardening modulus is defined in Equation (8) in terms of the elastic modulus \( E \) and input tangent modulus \( E_t \):

$$E_p = \frac{E \sigma_0}{E - E_t}$$

Figure 3 — Impact test configuration.
Failure is assumed to occur if

\[ \varepsilon_{\text{eff}}^p > \varepsilon_{\text{max}}^p \]

where \( \varepsilon_{\text{max}}^p \) is a user-defined input. In the model used for the bone, the effective failure strain was used as the \( \varepsilon_{\text{max}}^p \) criterion. Once failure has occurred, the deviatoric components are set to zero; that is, \( s_{ij} = 0 \) for all time.

To remain conservative, the failure strain used in this model was assumed to be at the low end of the range of ultimate strains for the cortical and cancellous bone found in literature (1.0–2.9%) (McCalden et al., 1993; McElhaney, 1966). In addition the failure strain of one plated sample \( P_2 \) was found experimentally to be 0.9%. The dynamic compression plate and screws were assigned the material properties of AISI 1010 Steel.

### Results

The load deflection curve for each sample is shown in Figure 4. To propagate crack growth while staying below the fracture instability point, the unplated samples were loaded to an average 450 N while the plated ones were loaded to an average 540 N. The actual critical load predicted by LEFM varied among samples. In Figure 4, the plated bone \( P_2 \) represents the plated bone with the LC-DCP plate, which was loaded to failure. The load deflection curves for the plated bones \( P_1, P_3 \) (DCP plates) exhibit nonlinearity around 400 N. The load deflection curves for the unplated bones \( U_1, U_2, U_3 \) appear to maintain linearity up to the critical load. The stiffness of the plated bones \( (P_1, P_3) \) are counterintuitively lower than the stiffnesses of the unplated bones \( (U_1, U_3) \).

Impact test results are shown in Table 4. The average value of the impact energy required to cause failure was 10.53% greater for the plated set. High-speed camera images revealed clean breaks with no rebounds. Based on high-speed camera images as well as visual inspection after impact testing, the unplated set crack displacement appeared to be predominantly Mode I while the dominant crack displacement mode for the plated set appeared to be Mode II. Figures 5 and 6 show representative sample fractures for the unplated and plated samples respectively.

The dynamic FE model was analyzed under different impact loads to study the effect of the plate on the stress distributions in the bone after impact. The variation of the maximum shear stresses, maximum normal stresses and shear strain along both bone configurations was investigated during and after failure. The maximum shear stress distributions at specific time steps during impact for the unplated bone and plated bones are shown in Figures 7

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**Figure 4** — Load deflection curve for all ulna samples subjected to static three-point bend \( (P_i: \text{plated ulnas, i.e., with attached fixation; } U_i: \text{unplated ulnas, i.e., without attached fixation}) \). Sample \( P_2 \): LC-DCP plated bone which was tested until failure. All other specimens were loaded up to crack growth instability point as predicted by LEFM.
and 8 respectively. It is observed from these results that the unplated bone has a maximum shear stress distributed within a 5 mm radius from the impact site, while the corresponding region is substantially larger, ∼100 mm, for the plated bone. Visual observation indicates that Mode II crack displacement appears to be dominant in the impact failure of the unplated set, while Mode I crack displacement appears to be dominant in the impact failure of the plated set.

**Discussion**

The LEFM coupled with composite beam theory predicted lower loads for damage occurrence within the
plated bones for the quasi-static loading case. This may be attributed to the plate being treated as completely fixed to the bone (i.e., no relative movement between plate and bone interface) which would result in larger strain gradients away from the plates (compared with the equivalent location in the unplated bones). For the plated bones, the LEFM predicted loads were close (within 15%) to the loads that would be predicted by 5% secant line as defined in ASTM E-399 for the critical load used to calculate fracture toughness for the three-point bend configuration. However, this was not the case for the unplated bones, leaving uncertainty as to whether new damage actually occurred before stopping the static test. The fact that the bones have an average age of 83.7 years allows a reasonable assumption that preexisting damage was present and would be increased with the application of a pseudo-static load.

An increase in bending stiffness due to the plate was only seen in one plated sample (P2); the absence of a consistent stiffness increase due to the plates is likely due to the high degree of variability in stiffness of the bones used (ages 77–90) as well as the relatively small sample.
size (six specimens). Note that despite the stiffness trends, the energy for failure for the damaged plated bones was on average higher than the energy required for impact failure of the damaged unplated bones. This suggests that the transverse impact response of damaged bone may be unrelated or only weakly related to its bending stiffness.

Based on an inspection of Figures 5 and 6, the fracture surface appears larger and the magnitude of damage does appear greater for the plated bone. This could be attributed to the high stiffness of the plate relative to the bone and the existence of local stress concentrations due to the screw holes. The greater potential for microcracks near the screw holes in the plated bone may contribute to its damage being more widespread. Since the unplated failure was more localized, there is a likely trade-off between increased impact resistance and increased damage due to impact. Due to the small sample size of the experiment, it cannot be concluded that the presence of the 3.5 mm stainless steel DC plate necessarily causes a 10.53% increase in the energy required for impact failure of the damaged ulna; however, a larger fracture surface for the plated bone is corroborated by the dynamic FE model predictions if fracture is attributed to maximum shear stress.

Implementation of a strain-based failure criteria in a linear orthotropic finite element dynamic model has been shown to generate predictions for impact response of a “damaged” ulna with and without plates that qualitatively corroborate impact test results where LEFM is employed to define the critical load below which stable crack growth will be maintained. Note that the experiments were performed to interpret computational predictions and understand the efficacy of the FE model. Neither the experimental procedure nor the computational model accounted for any decrease in cortical thickness due to contact stresses at the bone–plate interface, relative movement between the plate and bone, any decreased bone density due to stress shielding or reduction in metal stiffness and/or strength due to fatigue. All of these phenomena, if incorporated would weaken the bone–plate system and increase the quantity of damage. A decreased cortical thickness for the plated bone would have resulted in a lower $t/R_m$ ratio resulting in a smaller fracture toughness value. The consideration of relative motion between the plate and bone would create greater strain gradients, which would possibly increase the occurrence of microcracks; however, this leaves room for more plasticity events. LEFM is less applicable as plasticity effects become more significant. The authors would recommend a nonlinear fracture mechanics approach if these phenomena were to be incorporated. The authors note herein that, the inferences and observations reported in this study are only applicable for transverse mid-diaphyseal impact to the ulna in which the diameter of the impacting device is approximately twice the thickness of the cortical bone.

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**References**


