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INTERFRAGMENTARY SURFACE AREA AS AN INDEX OF COMMINUTION SEVERITY IN CORTICAL BONE IMPACT

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Summary

A monotonic relationship is expected between energy absorption and fracture surface area generation for brittle solids, based on fracture mechanics principles. It was hypothesized that this relationship is demonstrable in bone, to the point that on a continuous scale, comminuted fractures created with specific levels of energy delivery could be discriminated from one another. Using bovine cortical bone segments in conjunction with digital image analysis of CT fracture data, the surface area freed by controlled impact fracture events was measured. The results demonstrated a statistically significant ($p < 0.0001$) difference in measured *de novo* surface area between three specimen groups, over a range of input energies from 0.423 to 0.702 J/g. Local material properties were also incorporated into these measurements via CT Hounsfield intensities. This study confirms that comminution severity of bone fractures can indeed be measured on a continuous scale, based on energy absorption. This lays a foundation for similar assessments in human injuries.

Introduction

Bone fracture classification systems are the conventional means for clinically stratifying fracture severity, but they are not well suited for distinguishing among situations involving comminution. Part of this shortcoming no doubt arises from the intrinsic subjectivity in assessing severity of comminution as a categorical variable [18]. Fracture severity occurs on a continuum, and it should therefore ideally be measured on a continuum.

For several decades, fracture mechanics principles have been utilized to examine various aspects of bone failure [14]. The present study extends these principles to quantify comminution. Although never explicitly demonstrated, the intuitive relationship between degree of comminution and energy absorption is widely cited throughout the orthopaedic trauma literature. Experienced clinicians generally ascribe injuries with large numbers of fragments to “high energy” accidents. This is corroborated from fracture mechanics principles, which outline a direct correlation — in brittle materials — between the energy delivered in crack propagation and the area of new surface liberated.

Computed tomography (CT) images, nowadays routinely a component of the standard of care in major orthopaedic trauma, provide the opportunity for quantitative analysis of interfragmentary surface area on a continuous scale. Inference of energy absorption from measured interfragmentary surface area therefore affords the basis for objectively quantifying comminution. Toward that goal, impactions were conducted on excised bovine cortical bone

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segments. Using the same clinical CT protocol to image the specimens before and after impact, we tested the hypotheses that fragment sets from replicate impacts having similar energy absorption would have similar interfragmentary area, and that *de novo* fragment surface area would increase linearly with energy absorption [20].

Methods

Thirty-six 70-mm long segments were cut from the diaphyses of fresh-frozen tibiae, harvested from skeletally mature cows. The ends of these segments were then milled to produce two transverse parallel faces. CT scans of the bone segments were acquired prior to impactation using a standard orthopaedic protocol (a helical scan at 235 mAs and 140 kVp, 1 mm increment x 2 mm slice thickness x 1mm reconstruction, 0.25 mm in-plane pixel size (*i.e.*, a 128 mm field of view with a 512 x 512 pixel matrix)). The total surface area of the transverse end faces and of the endosteal and periosteal surfaces was computed for each intact specimen, using digital image analysis [3]. (Henceforth, surface area of the intact specimen will be referred to as “tare surface area”). The CT data were scrutinized for any obvious cracks or flaws in the specimens.

Using an instrumented drop tower (Figure 1), testing was then conducted at one of three distinct energy density levels (low = 0.423, middle = 0.532, and high = 0.702 J/g (n = 12 each)). The specimens were maintained in wet condition during the experiment. At each energy level, the same impactor mass was used (3.4, 6.0, and 7.7 kg respectively); drop height was adjusted based on specimen weight, in order to achieve parity of energy delivery per unit mass within groups. (Note that because density is fairly constant in bovine cortical bone, this is approximately tantamount to delivering a uniform energy per unit specimen *volume* within each group). The corresponding impact velocity utilized was approximately 5.2 ± 0.2 m/s, and did not differ significantly between the three energy groups.

All fragments were collected post-impact. They were then suspended in a specially prepared resin (Wood Epox[®], Abatron, Inc., Kenosha, WI), to which barium sulfate had been added to mimic the approximate CT density of soft tissue (+10 to +60 Hounsfield Units (HU)). Helical CT scans of these preparations were collected, using the same standard orthopaedic protocol as used on the intact bones. Thirteen specimens were scanned on a Toshiba XS (Toshiba, Tustin, CA), and twenty-three with a Marconi Mx8000 multi-slice scanner (Marconi Medical Systems Inc., Highland Heights, OH). In each case, phantoms (the Toshiba Medical Phantom and the Mx8000 System Phantom (serial #9178), respectively) were included to ensure consistency between scans/scanners, and scan parameters were replicated as closely as possible between the two machines.

Surface area measurements were extracted from the CT data, slice by slice for each fragment, using a digital image analysis algorithm developed expressly for this purpose [3]. These area values were summed across all fragments, and the original (tare) surface area of the intact specimen was subtracted. The fragment size distributions (including new and original surface) were plotted for each of the groups. Also, the bone surface area liberated per unit input energy (*i.e.*, the energy-to-surface conversion factor) was calculated for each specimen. *De novo* surface area was compared using an ANOVA, to test the hypothesis that the liberated surface area was greater in the specimens subjected to higher energy impacts. An ANOVA was also used to test for differences in surface area production per unit energy absorption, and for differences in fragment count. When statistically significant differences were found with ANOVA, *post hoc* group-to-group comparisons were made using Tukey's test.

Fracture toughness parameters are material-dependent, and are known to vary in a prescribed way with apparent bone density [4,13,21]. To account for heterogeneity of the bone material across specimens, calculation of a Hounsfield-based fracture energy measure was integrated

into the image analysis algorithm. Specimen-specific bone density values were obtained by first calculating the median Hounsfield intensity of all pixels previously identified as bone in the course of surface area calculations. The apparent wet bone density was then regressed using a linear relationship previously published in the literature [6]. From fracture mechanics theory [7], the energy required to liberate the fracture surfaces can be related to the apparent density through multiplication by a single material constant (4440 N/m, derived from experiments), once the densities are normalized to the density of the base bone material (2.4 g/cc, based on observed Hounsfield values in the bovine specimens). This approach is appropriate for bone of one type only (i.e., cortical-only bovine data).

Results

The thirty-six pre-fracture specimens ranged in weight from 65 to 149 grams (120.4 ± 17.6 grams). The initial surface area of the bones, i.e., the full endosteal and periosteal surfaces plus the two machined transverse faces, was $17,243 \text{ mm}^2 (\pm 2,360)$. No cracks were visible in any of the pre-testing specimen CT images.

Bovine segments impacted with 0.423 J/g (low-energy group) tended to exhibit distinctly different fragmentation patterns than those that absorbed 0.532 (middle-energy group) or 0.702 J/g (high-energy group) (Figure 2). Fragment size distribution plots (Figure 3) showed that in the high-energy group, the majority (65%) of surface area was comprised of fragments that each contributed less than 10% of the total specimen surface area. Conversely, 79% of the surface area in the low-energy group was contributed by large fragments (fragments that singly constituted over 30% of the total specimen surface area). The middle-energy group fell between these two, with 16% of the surface area resulting from the largest fragments, and 40% resulting from the smallest fragments. The fragment size distributions for the three groups of fragment sets followed the same principles of comminution that have been observed in other materials. That is, higher energy absorption produced a greater number of small fragments. The average number of fragments rose from 20.2 ± 10.9 , to 85.5 ± 48.5 , to 116.9 ± 62.4 , from the low to high-energy group. Likewise, the average fragment size tended to decrease, with an accompanying reduction in the standard deviation, when energy absorption increased (low = $995 \pm 3266 \text{ mm}^2$, middle = $393 \pm 1447 \text{ mm}^2$, and high = $332 \pm 1016 \text{ mm}^2$ of surface area).

A statistically significant difference in fragment count was found between groups with ANOVA, but closer scrutiny with a Tukey's *post hoc* test revealed that fragment counts for the middle and high-energy groups were not significantly different from one another ($p = 0.098$). While the middle and high-energy groups were also somewhat close in terms of liberated surface area, the p value for area was much stronger ($p < 0.01$).

The quantity of *de novo* surface area (Table I) generated in the specimens that absorbed the greatest energy was significantly higher ($p < 0.0001$) than the *de novo* area in each of the lower energy groups. Unexpected in principle from engineering fracture mechanics, the energy-to-surface conversion factor differed between the three groups ($p = 0.004$; $p > 0.05$ only between middle and high-energy groups in *post hoc* comparisons). A linear trend ($R^2=0.66$) between liberated surface area and energy absorbed in fracture was evident (Figure 4).

The data presented in Figures 3 and 4, and in Table I, are based on surface area measurements made without regard for any density variation in the cortical bone. The results of incorporating density, extrapolated from CT Hounsfield values, into a fracture energy estimate are shown in Figure 5. Specimen-specific bone densities in the bovine segments averaged $1.91 (\pm 0.10) \text{ g/cc}$. Incorporating this density correction did not noticeably improve data dispersion in the highest energy group, but it did improve separation between data in the middle and high-energy groups. Hounsfield values were normalized to a value of 2000 HU, corresponding to a density

of roughly 2.4 g/cc. This normalization value was selected as a liberal estimate of the maximum Hounsfield density of bovine cortical bone, in the absence of pathology. A variety of maximum Hounsfield values were provisionally utilized for normalization, and it was found that the choice of this value did not appreciably influence trends in the results.

Discussion

As one would expect, there was considerable scatter in our bovine impact data (both liberated surface area and efficiency of surface production), substantially more than in analogous studies [3] conducted using homogeneous and geometrically regular bone surrogate specimens. Among efforts to minimize scatter, parallel transverse faces were machined on the bovine tibial segments to facilitate uniform alignment in the drop tower chamber. Using transcortical specimens, rather than whole bovine tibiae, also aimed to reduce the influence of complex and variable geometry on impact testing of biological structures. Incorporation of bone density into a fracture energy measure helped to reduce the intra-group variability, and it helped specifically to reduce intergroup overlap between the medium and high-energy groups.

Bovine test energy dosages were delivered in drop tower experiments on a per-gram basis, in lieu of delivery per unit volume. Rabl and colleagues [17] found that volume-specific weight of human tibial specimens was inversely proportional to age. However, many slaughterhouse cow bones, perhaps due to the animals' similar age, diet, and exercise regime, share nearly uniform bone density and, therefore, volume-specific weight.

Despite scatter, linear regression demonstrated that bovine fracture specimens do indeed follow fracture mechanics principles in terms of a linear trend of energy versus surface area ($R^2 = 0.66$ for data in aggregate). The regression line, if extrapolated, crosses the abscissa at a distinctly nonzero value (Figure 4); this is consistent with other mechanisms of energy absorption also being operative. This consideration might perhaps explain the apparent variation in the energy-to-surface conversion factor between the low and high-energy groups. Fragment count, a factor sometimes incorporated into traditional dichotomous fracture classification schemes, was not nearly as good an indicator of fracture severity (energy absorption) as was liberated surface area. The bone fragment size distributions also varied as expected, with an effective decrease in mean fragment size with increasing energy absorption.

Because the constant of proportionality between energy delivery and liberated surface area is a material-dependent property [1], application of this paradigm to bone necessitated incorporating bone quality (*i.e.*, density). Intrinsic biologic variability in the mechanical fracture properties of bone is seemingly more of an issue in humans than in cattle, as there is much more variation in diet, genetics, and lifestyle (and consequently bone density) among people than among farm animals. When osteoporosis is present in elderly human bone specimens, in extreme cases, the bone surface can sometimes actually be compressed even with finger pressure [11].

The spatial distribution of the material in a bone will govern how well it resists crack propagation due to mechanical stresses. To that end, the correlation between apparent bone density and bone fracture toughness is well established [4,13,21]. Information regarding local bone apparent density at fracture sites is routinely available as CT Hounsfield numbers, which were incorporated into the present image analysis calculations. On the Hounsfield scale, the x-ray absorption of water has an assigned value of zero, whereas the x-ray absorption of air is negative one thousand. Over the range of densities found in biological tissues, for all practical purposes, the Hounsfield value varies linearly with tissue density [6,9,10,15]. Minor variability can exist in this relationship from raw scan to raw scan, but compensatory adjustments can be derived directly by placing a phantom of known density in the field.

One potential cause of difference between bovine versus human bone fracture propensity is the former's plexiform structure. Plexiform bone is orthotropic, and possesses superior stiffness and ultimate strength as compared to remodeled Haversian bone [12]. However, Haversian remodeling produces smaller and more nearly circular osteons which increases fracture toughness [5], since osteonal structure inhibits crack growth [8]. These two interrelated effects tend to cancel one another out, and Norman *et al.* [16] have shown that there is an approximate equality of these two species' fracture toughnesses, relative to their respective strengths. In the present series, no adjustments were made for Haversian versus plexiform character.

Microcracking during macroscopic crack propagation is one mechanism contributing to bone's fracture toughness [19]. The formation of microcracks, and their orientation, is ultimately linked to the microstructure of cortical bone [22], which differs between plexiform bovine bone and Haversian human bone. Vashishth *et al.* [19] noted that human bone fracture toughness specimens contained microcracks that were predominantly longitudinal (90% of cracks), whereas bovine bones from the same experiment contained 44% longitudinal microcracks, 44% inclined microcracks, and 12% transverse microcracks. In the present study, only macroscopic cracking was incorporated in the energy analysis.

In summary, the results of the current study demonstrate, for the first time, that fracture severity for impacted cortical bone can be characterized on a continuous scale, based on CT-apparent energy absorption. Obviously, utilizing this energy relationship in the clinical domain will involve dealing with a number of complexities that were absent from these tests of isolated bone segments. Also, the clinical severity of a comminuted fracture injury involves important considerations of soft tissue damage, in addition to the bony fracture *per se*. Nevertheless, this fragmentation energy approach introduces a physically justified basis for quantification in an area which has been until now purely the domain of subjectivity.

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Figure 1.

The bovine tibial segments were tested in a custom-designed and built drop tower which allowed quantification of the energy absorbed in fracture of the specimen.

Figure 2. Delimited fragments in typical CT slices, progressing left to right from lowest to highest energy, show variation in the patterns of fragmentation.

Figure 3.

Fragment size distribution plot for the thirty-four fractured bovine specimens. An initially overlooked large vessel foramen present in one specimen led us to exclude it from analysis. Data from a second specimen, identified as a statistical outlier by Grubbs' (ESD) test [2] (liberated surface area 2.4 standard deviations from the group mean), were also excluded from study. The "low energy" group (n=11) was nominally 0.423 J/g; "middle energy" (n=12) refers to 0.532 J/g; and "high energy" (n=11) was 0.702 J/g.

Figure 4.
Liberated surface area as a function of energy absorbed in fracturing the bovine bone drop tower series.

Figure 5. Bovine impact data with Hounsfield normalization incorporated to yield a fracture energy measure.

Table I

Liberated surface area (not adjusted for Hounsfield level) and energy-to-surface conversion in impacted bovine tibial cortical bone segments.

Energy Input (J/g)	Liberated Surface Area (mm ²)	Energy-to-Surface Conversion (mm ² per J)
0.423 ± 0.009	5232 ± 4410	122.6 ± 102.6
0.532 ± 0.012	14552 ± 4191	201.6 ± 56.8
0.702 ± 0.026	21757 ± 6708	243.2 ± 70.0