Published in final edited form as:

J Mech Behav Biomed Mater. 2011 October ; 4(7): 1021–1032. doi:10.1016/j.jmbbm.2011.03.011.

THE EFFECT OF STRAIN RATE ON FRACTURE TOUGHNESS OF HUMAN CORTICAL BONE: A FINITE ELEMENT STUDY

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Abstract

Evaluating the mechanical response of bone under high loading rates is crucial to understanding fractures in traumatic accidents or falls. In the current study, a computational approach based on cohesive finite element modeling was employed to evaluate the effect of strain rate on fracture toughness of human cortical bone. Two-dimensional compact tension specimen models were simulated to evaluate the change in initiation and propagation fracture toughness with increasing strain rate (range: 0.08 to 18 s⁻¹). In addition, the effect of porosity in combination with strain rate was assessed using three-dimensional models of microcomputed tomography-based compact tension specimens. The simulation results showed that bone's resistance against the propagation of fracture decreased sharply with increase in strain rates up to 1 s⁻¹ and attained an almost constant value for strain rates larger than 1 s⁻¹. On the other hand, initiation fracture toughness exhibited a more gradual decrease throughout the strain rates. There was a significant positive correlation between the experimentally measured number of microcracks and the fracture toughness found in the simulations. Furthermore, the simulation results showed that the amount of porosity did not affect the way initiation fracture toughness decreased with increasing strain rates, whereas it exacerbated the same strain rate effect when propagation fracture toughness was considered. These results suggest that strain rates associated with falls lead to a dramatic reduction in bone's resistance against crack propagation. The compromised fracture resistance of bone at loads exceeding normal activities indicates a sharp reduction and/or absence of toughening mechanisms in bone during high strain conditions associated with traumatic fracture.

Keywords

Cohesive finite element method; Fracture toughness; Strain Rate; Cortical bone

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1. Introduction

Bone is subject to a wide range of strain rates during daily activities such as walking (0.004 s^{-1}) (Lanyon et al., 1975), strenuous activities such as sprinting and downhill running (0.05 s^{-1}) (Burr et al., 1996) or traumatic fracture events such as accidents or falls (25 s^{-1}) (Hansen et al., 2008). Previous studies showed that the mechanical response of bone, including its modulus of elasticity, yield stress and strain, and ultimate stress and strain vary with the loading rate (McElhaney and Byars, 1965; Crowninshield and Pope, 1974; Saha and Hayes, 1974; Currey, 1975; Saha and Hayes, 1976; Wright and Hayes, 1976; Robertson and Smith, 1978; Evans et al., 1992; Hansen et al., 2008; Zioupos et al., 2008).

A comprehensive understanding of traumatic fractures requires an investigation of bone's resistance to fracture initiation and propagation under a variety of low and high strain rates. Most of the fracture toughness measurements reported in the literature under varying strain rates, however, corresponded to quasi-static conditions and reported only on fracture initiation. For example, using quasi-static conditions, some of the earlier studies done on bovine or equine bone reported an increase in initiation fracture resistance measured as energy absorption or critical stress intensity factor with increasing strain rates up to a certain level after which a decrease was observed (Piekarski, 1970; Crowninshield and Pope, 1974; Robertson and Smith, 1978; Behiri and Bonfield, 1980, 1984; Evans et al., 1992). More recent investigations also reported similar trends where fracture toughness in bovine and equine bone (Adharapurapu et al., 2006; Charoenphan and Polchai, 2007; Kulin et al., 2008; Kulin et al., 2010; Kulin et al., 2011) and energy to fracture in human cortical bone (Zioupos et al., 2008) decreased with increasing strain rate. The only study that measured the propagation toughness at a high strain rate reported the loading rate effects on the R-curve behavior of equine cortical bone (Kulin et al., 2010). This study showed that although the bone exhibited increasing fracture toughness with crack propagation under both quasi-static and dynamic loading, the propagation toughness was lower in the dynamic loading compared to the quasi-static case. Furthermore, the reduction in the propagation toughness was more pronounced compared to the reduction in initiation toughness (Kulin et al., 2010).

It is well known that the energy absorption during fracture of bone depends critically onto whether the fracture is stable (high energy) or unstable (low energy) with the fracture scenario mostly passing from a state of (i) diffuse widespread microcracking damage, to (ii) a localized stable crack growth, to (iii) an unstable growth of a crack front (Zioupos, 1998). The degree to which bone is brittle or tough depends on its ability to avoid a ductile-to brittle transition for as long as possible during the deformation between stages (i+ii) and (iii). With regard to strain rate effects in particular the key to bone's brittleness is the strain and damage localization early on in the process, which leads to low post-yield strains and low energy absorption to failure between stages (i) and (ii) (Zioupos et al., 2008). Crack growth behavior in stages (ii) and (iii) under different strain rates has been investigated by Charoenphan and Polchai (2007) who showed that the stress intensity factor in bovine bone increased with crack growth up to the point of unstable crack growth, after which the values started to decrease. In addition to experimental studies on cortical bone, finite element models incorporating varying strain rates have been also developed to predict the experimental measurements of energy release rate in cortical bone (Charoenphan and Polchai, 2007), and the effect of variation of architecture and strain rates on trabecular bone fracture behavior (Tomar, 2008).

The review of the literature shows that the experimental studies in the literature mostly reported the measurement of initiation fracture toughness or energy to fracture but did not focus on propagation fracture toughness with the exception of one study (Kulin et al., 2010). Initiation and propagation toughness represent different fracture processes in bone

(Vashishth, 2004). Bone exhibits increasing fracture toughness with crack propagation following crack initiation (Vashishth et al., 1997; Nalla et al., 2005). Physiological everyday loading inherently leads to the creation of in-vivo microcracks about 50–100µm long, which accumulate with age increasing in number and density due to the fact that the human body is less able to repair them in later life (Schaffler et al., 1995; Norman and Wang, 1997; Zioupos, 2001). Such observations indicate that the propagation toughness is the critical measure of toughness that evaluates the crack growth resistance in bone, and this has to encompass relevant conditions such as variable strain rate of loading, aging, increasing porosity and so forth.

The overall goal of the current study is to develop a computational approach that evaluates the effect of strain rate on both initiation and propagation toughness of human cortical bone. The experimental study (Kulin et al., 2010) conducted on propagation toughness measurement only focused on two strain rates that represent quasi-static and dynamic loading and did not investigate a wide range of strain rates. Furthermore, the experiments were carried out using equine cortical bone. The current study will provide additional information on the response of human cortical bone under various strain rates ranging between 0.08 to 18 s⁻¹. The computational method used in this study is cohesive finite element modeling which has been applied and validated previously in other applications (Ural and Vashishth, 2006b, 2007b, a). Unlike experimental studies, the use of a computational approach enables controlled evaluation of the effects of a single parameter on the fracture response of bone. This feature of computational modeling ensures that there are no additional confounding factors (such as those present in an experimental approach i.e inter- and intra-donor variation) included in the evaluation. In the current study, using two-dimensional simulations of compact tension specimen geometry, we varied the strain rate while keeping the same specimen geometry, material properties and microstructure to determine the effects of strain rate on the initiation and propagation fracture toughness of human cortical bone. The measured fracture toughness values were tested for correlations with the amount of microdamage that bone accumulates in a diffuse/microcracking manner when tested over a range of strain rates in the absence of a major notch (Zioupos et al., 2008). Furthermore, using three-dimensional models of microcomputed tomography-based compact tension specimens (which provided realistic structures with fine micro detail) the effect of porosity was also assessed in combination with strain rate to determine the influence of microstructure on fracture behavior under varying strain rates. This aims to improve our current understanding of fall-related fractures in the elderly.

In summary, the current study focuses on determining (a) the variation of initiation and propagation fracture toughness with strain rate (b) the correlation of microcracking with initiation and propagation fracture toughness (c) the combined effect of porosity and strain rate on initiation and propagation toughness.

2. Methods

The effect of strain rate on fracture toughness of bone was investigated via compact tension specimen models which were created using the finite element program, ABAQUS (version 6.8, 2008, Simulia, Providence, RI). The models that were investigated included a two-dimensional (2D) model of a compact tension specimen used in previous experimental studies (Vashishth et al., 1997; Vashishth et al., 2004), and a three-dimensional (3D) compact tension specimen model obtained via micro-computed tomography (µCT) (Ural and Vashishth, 2007b). The 2D model was used to evaluate the variation of initiation and propagation fracture toughness with strain rate and the 3D µCT-based model was utilized to investigate the interaction of porosity with varying strain rate.

2.1. Experimental Data

The simulations performed in this study utilized experimental data (Table 1) that evaluated the mechanical properties of bone at various strain rates (Hansen et al., 2008; Zioupos et al., 2008). These studies used dog-bone shaped tensile specimens of cortical bone machined parallel to the longitudinal axis of the human femur. The specimens were tested under five displacement rates, namely, 1, 10, 50, 100 and 200 mm/s, creating strain rates between 0.08 s^{-1} and 18 s^{-1} . The experimental measurements were done on 25 specimens including four to six specimens for each of the five displacement rates. The measurements from the tests showed that with an increase in strain rate, the tensile elastic modulus increased linearly, the ultimate tensile strength decreased exponentially. In addition, while the overall fracture energy and post-yield energy to fracture per unit volume showed a significant decrease with increasing strain rate (Figure 1a and 1b), the energy to fracture beyond the maximum stress was almost constant (Figure 1a) (Hansen et al., 2008; Zioupos et al., 2008). The comparison of the experimental data used in this study to previous work is shown in Figure 1b. As this comparison shows, the current study focuses on the strain rate regime after the transition from increasing to decreasing trends in energy to fracture. Furthermore, microcracking in the tested specimens was evaluated by adding the assigned numerical scores (from 0 (no microcracking) to 3 (widespread microcracking)) at two different locations in the specimens. The microcracking scores showed an exponential decrease as the strain rate increased (Zioupos et al., 2008). These measurements provide a set of material properties that can be used in finite element simulations to simulate the effect of strain rate on fracture toughness of bone.

2.2. Computational Model

The simulations performed in the current study utilized cohesive finite element modeling which has been previously applied to fracture toughness evaluation of bone (Ural and Vashishth, 2006b, 2007b, a). Cohesive finite element modeling is a nonlinear fracture mechanics approach that represents the physical processes occurring in the vicinity of a propagating crack by a simplified traction-crack opening displacement relationship (Figure 2d). The traction-displacement curve, defining the cohesive model, is composed of ascending and descending branches that incorporate material softening and nonlinearity. The model isolates the fracture process from the surrounding continuum constitutive model and represents the nonlinear behavior of the process zone, which cannot be captured by linear elastic fracture mechanics analysis. Cohesive model is a phenomenological representation of all the fracture mechanisms (such as crack bridging, and microcracking) as an overall effect in the vicinity of the crack and does not attempt to explicitly resolve each mechanism.

Cohesive models are defined by a traction-displacement relationship based on three parameters, two of which uniquely define the fracture process. These parameters are the critical strength of the material (σ_c), a characteristic crack opening displacement at fracture (δ_u), and the energy needed for opening the crack (area under the traction-displacement curve, G_c) (Figure 2d). The cohesive parameters can be obtained from macroscale tests representing the stress-displacement relationship in the damage zone (Gdoutos, 2005). A typical stress-strain graph for bone is shown in Figure 2a as well as experimentally measured stress-strain curves under different strain rates used in the current study (Figure 2b). The damage accumulation and microcrack formation are marked by the nonlinear increasing and decreasing part of the curve. The energy associated with these regions of the stress-strain diagram corresponds to the local energy dissipation due to damage formation (Figure 2c). As a result, the energy to fracture in the stress-strain curve can be correlated to the fracture behavior. In addition, previous studies on the prediction of fracture toughness of bone using properties measured in macroscale experiments showed good agreement with experimental fracture toughness data (Ural and Vashishth, 2006b) and demonstrated the

predictive capability of the modeling approach. Previous studies showed that the most influential cohesive model parameters that affect the fracture behavior are the fracture toughness and the strength whereas the shape of the cohesive relationship has little effect on the resulting fracture behavior (Tvergaard and Hutchinson, 1992). Therefore, in this study, we employed a simple bilinear traction-displacement curve of the following form:

$$T = \begin{cases} \frac{\sigma_c}{\delta_c} \delta & \delta \leq \delta_c \\ \sigma_c \frac{(\delta_u - \delta)}{(\delta_u - \delta_c)} & \delta_c < \delta \leq \delta_u \\ 0 & \delta \geq \delta_u \end{cases} \quad (1)$$

where T and δ are the scalar effective traction and opening displacement representing the effects of the normal and shear components; σ_c is the peak traction representing the maximum strength of the cohesive model; δ_c is the critical displacement for crack initiation; and δ_u is the failure displacement related to the formation of a complete crack. The critical displacement (δ_c), where the crack initiates, is not a physical quantity and is chosen as the smallest value satisfying the numerical convergence.

In the finite element context, cohesive models are represented as line elements or surface elements with zero thickness in 2D and 3D, respectively, and are compatible with solid elements (Figure 2e and 2f). Each cohesive element follows the traction-crack opening profile described above and forms a crack when there is no transfer of traction between the opening surfaces (i.e. $T = 0$ and $\delta = \delta_u$). The details of the application of cohesive modeling to cortical bone can be found in Ural and Vashishth (2006, 2007a,b).

2.3. Finite Element Modeling of 2D Compact Tension Specimens

Two-dimensional compact tension specimens were created using ABAQUS based on the specimen size in experimental studies on cortical bone (Vashishth et al., 1997; Vashishth et al., 2004) (Figure 3a). The models were composed of 6638 quadrilateral elements and contained 368 cohesive elements (Figure 3b). The number of cohesive elements was chosen to satisfy the mesh convergence criteria reported in a previous study (Ural and Vashishth, 2006b). The cohesive interface elements are placed in the direction of the expected crack growth as shown in Figure 3b. The displacement boundary conditions were applied at the center of the holes and were increased incrementally. The model was fixed at the midpoint of the farthest edge from the loading (Figure 3b).

The material properties assigned to the model were based on the experimental data, which are briefly summarized in Section 2.1 and reported elsewhere in detail (Hansen et al., 2008; Zioupos et al., 2008). The elastic modulus and the cohesive model properties (G_c and σ_c) were varied based on 25 experimental measurements at strain rates between 0.08 to 18 s⁻¹. The elastic modulus was used for defining the behavior of the bulk material. The continuum elements were assigned linear elastic isotropic material properties. Although the bone has anisotropic properties, this simplification was introduced in the models due to the availability of the elastic modulus variation with strain rate only in the longitudinal direction (Hansen et al., 2008). Cohesive model properties, including the critical strength (σ_c) and energy release rate (G_c), were also varied with respect to the strain rate. Ultimate tensile strength measured in the experiments was used as the critical strength (σ_c) of the cohesive model. Energy release rate (G_c) was taken to be proportional to the energy to fracture that was reported in each of the experiments. The simulations were carried out under plane strain assumptions and were run as quasi-static simulations. The strain rate effects were captured through the variation of the material properties as a function of strain rate.

The load and crack growth data computed in the 25 simulations were used to calculate the crack growth resistance (K_R) at various crack lengths using the equation (ASTM-E399-90, 2000)

$$K_R = \frac{P}{BW^{1/2}} f\left(\frac{a}{W}\right)$$

$$f\left(\frac{a}{W}\right) = \frac{(2 + \frac{a}{W})(0.886 + 4.64\frac{a}{W} - 13.32(\frac{a}{W})^2 + 14.72(\frac{a}{W})^3 - 5.6(\frac{a}{W})^4)}{(1 - \frac{a}{W})^{3/2}} \quad (2)$$

where P is the load, a is the crack length, W is the width of the specimen from the load line, and B is the thickness. For each simulation, K_R is plotted as a function of crack extension, (defined as R-curve) to obtain the resulting slope (defined as R-curve slope) by linear regression. The R-curve slope was calculated for crack extensions of up to 2.5 mm in the simulations. The initiation toughness (K_{init}) corresponded to the stress intensity factor at the first crack growth increment in each simulation.

2.4. Finite Element Modeling of 3D μ CT-Based Compact Tension Specimens

Three-dimensional μ CT-based models of compact tension specimens incorporating porosity were created to investigate the interaction of porosity and strain rate. The models used in the simulations were based on two previously tested compact tension specimens of human cortical bone (19- and 81-year-old) scanned using μ CT (vivaCT 40, Scanco Medical AG) (Figure 3c). The porosity of the specimens was computed using the 3D image evaluation capability provided by the μ CT software (μ CT Evaluation Program, version 5.0, vivaCT 40, Scanco Medical AG). The details of the creation of these models and porosity evaluation can be found in a previous finite element study (Ural and Vashishth, 2007b).

The finite element meshes consisted of linear cube-shaped elements and consisted of 84,541 and 66,601 elements for the young (19 years) and the old (81 years) donors, respectively (Figure 3d). In addition, 19-year-old specimen had 1% whereas 81-year-old specimen had 5% overall intracortical porosity. The models included no age-related material property changes and embodied only the difference in microarchitecture that existed between these two ages so as to identify simply the relationship between the strain rate and bone porosity. The loading, the boundary conditions and the location of the cohesive elements were the same as in 2D simulations. The material properties including the elastic modulus and the cohesive model properties assigned to the models were based on the experimental data reported in the previous sections. The simulations were run using the material properties obtained from 10 experimental measurements, which were a subset of the 25 strain rate measurements used in 2D simulations. These included representative experimental data from all displacement rates (1 to 200 mm/s). The simulation results were evaluated for initiation and propagation fracture toughness following the equations and procedures used for the 2D compact tension specimens.

3. Results

3.1. 2D Compact Tension Specimen Simulations

2D compact tension specimen simulations evaluated the change in initiation and propagation toughness with strain rate. The simulation results showed that both initiation and propagation toughness decreased with increasing strain rate (Figure 4). The initiation toughness and R-curve slope values found in the simulations varied between 7.8 to 28.4 MPa m^{1/2} and 0.63 to 5.7 MPa m^{1/2}/mm, respectively. The fracture toughness values obtained from the simulations are presented as normalized values with respect to the largest value (of all values observed in the low strain rate group) for initiation and propagation

toughness (Figure 5a and 5b) to highlight the percent changes in initiation and propagation toughness with strain rate. The simulation results showed that initiation toughness exhibited a nonlinear decline with an increase in strain rate (Figure 5a). On the other hand, R-Curve slope, denoting the propagation fracture toughness, showed a sharp decrease up to a strain rate of 1 s^{-1} and attained an almost constant value for strain rates $> 1 \text{ s}^{-1}$ (Figure 5b). R-curve slopes had significantly lower values at high strain rates that were about 20% of the highest fracture toughness value, whereas initiation fracture toughness did not decrease as drastically and was about 40% of the largest value at the highest strain rate evaluated (Figure 5a and 5b).

The variation of initiation and propagation fracture toughness with strain rate can be captured by an exponential equation of the form $y = a + be^{-c\dot{\epsilon}}$ where y denotes the initiation or propagation fracture toughness, $\dot{\epsilon}$ denotes the strain rate, and a , b , and c are the coefficients of the fit equation. The coefficients a , b , and c for the curve fits shown in Figure 5a and 5b are listed in Table 2. The exponential fits provided good correlations between the strain rate and initiation fracture toughness ($R^2 = 0.80$) as well as the propagation fracture toughness ($R^2 = 0.78$) (Table 2). Different fracture toughness values obtained for the same strain rate was due to the variation in the experimental material data that was used as input parameters in the simulations.

Normalized K_{init} and R-Curve slope values were also plotted as a function of the microcracking score (denoted as “microcrack number”) assigned to the tested specimens reported by Zioupos et al. (2008). Both initiation and propagation fracture toughness showed a positive linear correlation with microcrack number (Figure 6a and 6b). As the microcrack number increased, both initiation and propagation toughness also increased. The correlation values obtained for the linear fits were $R^2 = 0.65$ ($p < 0.001$) and $R^2 = 0.45$ ($p < 0.001$) for initiation and propagation toughness, respectively. These correlation values are better or similar to the values seen in the experiments of Zioupos et al. (2008) between ‘microcrack number’ vs. ‘strain at yield’ ($R^2 = 0.45$) and ‘microcrack number’ vs. ‘post-yield’ ($R^2 = 0.56$).

3.2. 3D μ CT-based Compact Tension Specimen Simulations

3D μ CT-based compact tension specimen simulations focused on evaluating the interaction between porosity and the strain rate. Similar to 2D models, the results showed that initiation and propagation toughness for both porosity levels decreased with increasing strain rates that can be best fit by an exponential curve in the form of $y = a + be^{-c\dot{\epsilon}}$. The coefficients a , b , and c for the curve fits shown in Figure 7 and 8 are listed in Table 3.

The initiation toughness and R-curve slope values varied between 4.6–16.2 $\text{MPa m}^{1/2}$ and 0.91–4.34 $\text{MPa m}^{1/2}/\text{mm}$, respectively for the specimen with lower porosity (19-year-old specimen). In addition, for the specimen with higher porosity (81-year-old specimen), the initiation toughness and R-curve slope varied between 3.1–10.3 $\text{MPa m}^{1/2}$ and 0.43–2.63 $\text{MPa m}^{1/2}/\text{mm}$, respectively. The variation in fracture toughness is reported as normalized values following two normalization schemes to highlight the percent changes in initiation and propagation toughness with strain rate. In the first approach, fracture toughness from both specimens were normalized with respect to the overall largest initiation and propagation toughness that was obtained for the model with lower porosity (19-year-old specimen) (Figure 7a and 8a). These results indicate that the initiation and propagation fracture toughness decrease with increasing porosity and with increasing strain rate.

In the second normalization approach, the initiation and propagation toughness were normalized with respect to their individual largest values (Figure 7b and 8b) and that is the best behavior these two structures showed at the lowest strain rates. This normalization

approach eliminated the influence of the porosity on the fracture toughness variation. For initiation fracture toughness, the curves for both specimens almost fell on the same curve (Figure 7b). The influence of the strain rate on initiation fracture toughness was the same irrespective of the porosity of the bone with no additional interaction of strain rate and porosity. However, for propagation fracture toughness, the model with higher porosity showed lower propagation fracture toughness for all strain rates (Figure 8b) highlighting the further reduction in propagation fracture toughness as a result of the interaction between strain rate and porosity.

4. Discussion

This paper investigated the effects of strain rate on fracture behavior of human cortical bone using a nonlinear fracture mechanics based cohesive finite element model (Ural and Vashishth, 2006b, 2007b, a). The simulations focused on determining the change in initiation and propagation toughness with strain rate via 2D and 3D μ CT-based compact tension specimens. The material properties used in the simulations were based on measurements performed on human cortical bone specimens under different strain rates (Hansen et al., 2008; Zioupos et al., 2008). The aim was to corroborate some results produced in fracture mechanics tests of the past (Behiri and Bonfield, 1980, 1984), even if they only apply for initiation fracture toughness, and if possible elucidate this fracture process.

This study reports predicted values of propagation toughness under a wide range of strain rates and provides new and unique information on the variation of propagation toughness under high impact loading conditions. In addition, the current study reports the effect of porosity on the mechanical properties at high strain rates, which has not been investigated before. Therefore, the results of this study provide a new insight into the fracture behavior of bone and its interaction with its microstructure during traumatic fracture incidents.

Under quasi-static conditions, studies of the initiation and propagation fracture toughness behavior of bone have revealed the very different processes that relate to these two measures (Vashishth, 2004). Initiation toughness reflects the propensity of bone to form microcracks. By contrast, propagation toughness relates to the crack growth resistance of bone resulting from various toughening mechanisms such as microcracking, uncracked ligament bridging, crack bridging and crack deflection (Vashishth et al., 1997; Nalla et al., 2003; Nalla et al., 2005; Koester et al., 2008).

Our study found that initiation and propagation fracture toughness exhibited differences under varying strain rates. The simulation results showed a sharp initial decrease in propagation toughness with strain rates between $0.08\text{--}1\text{ s}^{-1}$ down to 20% of its initial value, but no change in its value for strain rates $> 1\text{ s}^{-1}$, when presumably all crack growth is fast and stays in an unstable mode. On the other hand, initiation fracture toughness exhibited a more gradual decrease to 40% of its initial value over the strain rate range we have examined. These results show that the initiation fracture toughness changes much less with strain rate than propagation fracture toughness, and may be a less sensitive measure of the change in crack resistance of bone we see in traumatic conditions.

More importantly, in contrast to physiological loading where bone exhibited a substantial increase in crack growth resistance with crack propagation (i.e. rising R-curve with a large slope), high strain rate loading resulted in a much smaller R-curve slope in magnitude. This difference in R-curve slope values suggests a sharp reduction and/or absence of toughening mechanisms in bone during high strain rates such as these associated with traumatic fracture conditions. This agrees well with: (1) the smooth fracture surface profiles observed in

specimens tested under high loading rates (Behiri and Bonfield, 1980; Adharapurapu et al., 2006; Zioupos et al., 2006; Kulin et al., 2008); (2) the good positive correlations we have observed here between the experimentally measured microcrack number and the fracture toughness found in our simulations; (3) the previous study by Vashishth et al., (1997) which demonstrated that increased microcracking leads to higher bone toughness; (4) the experimental observations by Kulin et al. (2008, 2010, 2011) that bone subjected to quasi-static loading undergoes significant peripheral damage during crack propagation whereas bone subjected to dynamic loading did not exhibit a large amount of damage; (5) the observations by Zioupos et al. (2008) that the key to bone's brittleness in high strain rates is the strain and damage localization occurring early on in the process, which leads to low post-yield strains, low energy absorption to failure and less collateral damage alongside the fracture front.

The highest R-curve slope found in this study ($5.7 \text{ MPa m}^{1/2}/\text{mm}$) corresponding to the lowest strain rates are close to the quasi-static R-curve slope predictions reported in the literature ($5.93 \text{ MPa m}^{1/2}/\text{mm}$) (Kulin et al., 2010). On the other hand, the lowest R-curve slopes predicted under high strain rates ($0.63 \text{ MPa m}^{1/2}/\text{mm}$) were lower than the experimentally measured values ($3.81 \text{ MPa m}^{1/2}/\text{mm}$) by Kulin et al. (2010). These differences may stem from the differences in the equine and human cortical bone properties. Despite this discrepancy, the decreasing trend in the R-curve behavior that our simulations predicted agrees well with the experimental data. In addition, our predictions that showed a larger decrease in propagation toughness compared to initiation toughness also agrees with the experiments. Our simulations did not capture the transition from increasing to decreasing trend in fracture toughness reported in earlier studies (Behiri and Bonfield, 1980, 1984) due to the strain rate range investigated. In the current study, the lowest strain rate modeled is 0.08 s^{-1} which coincides with the range of strain rate values that the increasing trend transitions to a decreasing behavior in tensile loading ($0.01\text{--}0.1 \text{ s}^{-1}$) (Figure 1b). In the current study, we captured the dynamic effects through the constitutive law defining the cohesive model. Although inertial effects may be of importance at higher values of strain rate, the simulations are successful at capturing the overall behavior that is obtained in experimental studies.

The comparison of the simulation results to other quasi-static experimental data show that both the R-curve slope and the initiation toughness for the low strain rates were generally higher than the experimentally reported values especially when compared to the experiments that measured fracture toughness parallel to the osteons. The cohesive properties that were used in the current simulations were taken from tensile tests that are performed in the longitudinal direction, therefore, represent crack formation in the transverse direction. Significant differences between transverse and longitudinal crack growth was demonstrated in the literature where stress intensity factors were shown to increase to levels of $25 \text{ MPa m}^{1/2}$ within 0.5 mm (Koester et al., 2008) which are in the vicinity of the stress intensity factors that are reported in this study for the lowest strain rate case. The higher values may also be influenced by the fracture toughness values assigned to the cohesive models based on the energy to failure measurements. Although this may influence the absolute values of the initiation and propagation toughness, it does not affect the relative values of initiation and propagation toughness that are found in the simulations for each strain rate and shown in normalized form in Figures 5, 7 and 8.

The cohesive model parameters have direct influence on the predicted initiation and propagation toughness values. Both cohesive strength (σ_c) and fracture toughness influence the K_{init} value obtained in the simulations through the initiation of damage and formation of the initial crack extension. On the other hand, R-curve behavior is mostly influenced by the

fracture toughness (G_c) parameter in the cohesive model. A reduction in the cohesive strength and fracture toughness leads to lower values of K and R-curve slope.

Physiological strain rate measurements yielded strain rates ranging from 0.005 to 0.08 s^{-1} for animals (Rubin and Lanyon, 1982), and between 0.004 and 0.05 s^{-1} for humans (Lanyon et al., 1975; Burr et al., 1996) indicating an upper bound of 0.08 s^{-1} during regular activities. On the other hand, traumatic fracture events such as in-car accidents or falls lead to higher strain rates with an estimated upper bound of 25 s^{-1} (Hansen et al., 2008). Some studies have reported that above a certain strain rate value in the range of 0.01–0.1 s^{-1} in tensile loading (Crowninshield and Pope, 1974; Evans et al., 1992) or 0.1–1 s^{-1} in compressive loading (McElhaney and Byars, 1965), the increasing energy absorption behavior of bone assumes a decreasing trend. These strain rate values, where the behavior changes are close to the upper bound of physiological strain rates reported in the literature. These observations combined with the current simulation results may suggest that bone as a material has not evolved towards handling high impact loads and has an impaired fracture resistance at loads exceeding normal activities including walking and running.

The findings in the current study have certain implications for fractures caused by low energy falls in the elderly. There is a well documented decline in the crack growth resistance of human bone with age (Nalla et al., 2004; Vashishth et al., 2004). There is a possibility that the reduction with age may be combined with a significant reduction in propagation fracture toughness with strain rate (for rates just above the physiological range), and this may herald a higher risk for fracture for our elderly population even for low energy falls. Furthermore, the present simulation results showed that an increase in porosity exacerbates the decreasing effect of strain rate on propagation fracture toughness. As the porosity of bone verifiably increases with age (Yeni et al., 1997; Ural and Vashishth, 2006a), the crack growth resistance of bone is expected to be further compromised and to result in high incidence of fall related fractures in the elderly.

In summary, the effects of strain rate on both initiation and propagation fracture toughness of human cortical bone were investigated using a computational approach based on cohesive finite element modeling. Our simulation results show that strain rates associated with fall loading conditions lead to a dramatic reduction in bone's resistance against initiation and propagation of a fracture and this effect is exacerbated by increased porosity. The current study brings new insight into fracture behavior of bone under high loading rates, which may help understand bone fractures due to traumatic events such as accidents and falls in the elderly.

Acknowledgments

The authors acknowledge: NIH Grant AG20618 (DV); the use of Imaging Core in Center of Biotechnology and Interdisciplinary Studies at Rensselaer Polytechnic Institute; and the support provided to PZ by DA-CMT to study 'stress fractures in young army recruits'.

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Figure 1.

(a) Variation in energy to fracture with strain rate (Hansen et al. 2008). White circles show the fracture energy beyond the maximum load and green circles denote the post-yield fracture energy (i.e. the total fracture energy minus the pre-yield (elastic) energy). (b) Comparison of energy to fracture between the experimental data used in this study (open circles-up to the yield point; grey diamonds-up to the max load; solid triangles total energy to fracture) and previous work (McE: McElhaney (1966), human femur in compression; C +P : Crowningshield & Pope (1974), bovine tibia in tension; E : Evans et al. (1992), horse metacarpals in tension; UH+PZ: Hansen et al.(2008), human femur in tension). The black and green lines correspond to the measurements presented in Hansen et al. (2008) representing the overall fracture energy and the post-yield fracture energy, respectively.

Figure 2d

Figure 2e

Figure 2.

(a) Schematics of stress-displacement diagram for bone showing the possible unstable crack growth (USCG) and stable crack growth (STCG) paths after maximum stress level. (b) Experimental load-displacement diagrams obtained under different strain rates (Hansen et al., 2008). (c) Schematics of crack formation corresponding to the regions identified in (a). (d) Traction-displacement relationship defining the cohesive zone model. (e) Quadrilateral solid elements and the compatible 2D cohesive element with four nodes used in two-dimensional compact tension specimen simulations. (f) Hexahedral solid elements and the compatible 3D cohesive element with six nodes used in μ CT-based compact tension specimen simulations.

Figure 3a

Figure 3.

(a) Schematics of compact tension specimen that was used in the finite element simulations. Note that $a/W = 0.52$ at the initial crack configuration. (b) Two-dimensional finite element mesh of the compact tension specimen. Note that triangles on the midpoint of the right edge of the specimen indicate fixed boundary conditions and the arrows denote the location and direction of applied loading. (c) μCT scans of compact tension specimens for 19-year-old (1% porosity) and 81-year-old (5% porosity) donors. (d) Finite element meshes generated from μCT scans of compact tension specimens machined from 19- and 81-year-old donor bones and the associated crack planes tiled with cohesive elements.

Figure 4.

Representative stress intensity factor (K) versus crack extension (da) plots for three strain rate values ($0.08, 9.75, 18 \text{ s}^{-1}$) obtained from the finite element simulations for 2D compact tension specimens. The initiation toughness decreases as the strain rate increases. Note that the stress intensity factors as well as the R-curve slope for 0.08 s^{-1} are much higher compared to larger strain rate cases.

Figure 5.

(a) Normalized K_{init} vs. strain rate for 2D compact tension specimens. (b) Normalized R-curve slope vs. strain rate for 2D compact tension specimens. Note that the fracture toughness values are normalized with respect to the largest value of initiation and propagation toughness among all simulations. The regression lines are based on an exponential equation of the form $y = a + be^{-c \dot{\epsilon}}$ with $R^2 = 0.80$ and 0.78 for (a) and (b) respectively.

Figure 6.

(a) Normalized K_{init} vs. microcrack number. (b) Normalized R-curve slope vs. microcrack number. Microcrack number denotes the total microcracking score (0–6) assigned to the specimens following the method outline in Section 2.1. Note that the fracture toughness values are normalized with respect to the largest value of initiation and propagation toughness among all simulations.

Figure 7.

(a) Normalized K_{init} vs. strain rate for 19 and 81-year old specimens. Each specimen is normalized with respect to the maximum K_{init} value of 19-year-old specimen. (b) Normalized K_{init} vs. strain rate for 19 and 81-year old specimens. Each specimen is normalized with respect to their maximum K_{init} value. The regression lines are based on an exponential equation of the form $y = a + be^{-c \epsilon}$.

Figure 8.

(a) Normalized R-curve slope vs. strain rate for 19 and 81-year old specimens. Each specimen is normalized with respect to the maximum R-curve slope value of 19-year-old specimen. (b) Normalized R-curve slope vs. strain rate for 19 and 81-year old specimens. Each specimen is normalized with respect to their maximum R-curve slope value. The regression lines are based on an exponential equation of the form $y = a + be^{-c\dot{\epsilon}}$.

Table 1

Material properties measured under varying strain rates (Zioupos et al., 2008; Hansen et al., 2008) that are used in the finite element simulations.

Displacement rate (mm/s)	Strain Rate (1/s)	Elastic Modulus (MPa)	σ_c (MPa)
1	0.08	20400	132.6
1	0.08	19700	112.0
1	0.08	16060	119.3
1	0.08	16740	107.4
1	0.08	13700	115.1
10	1.24	17000	127.4
10	0.81	17000	118.9
10	0.83	17750	136.1
10	0.82	12530	153.8
10	0.84	13600	137.1
10	0.82	14140	136.7
50	2.93	19200	107.7
50	3.42	19000	99.2
50	4.26	12800	142.6
50	3.52	17400	105.1
100	4.81	17000	80.8
100	4.34	18400	72.8
100	7.62	17000	64.2
100	10.31	18340	83.5
100	9.75	14280	87.5
200	17.86	20280	43.5
200	17.29	25800	45.7
200	17.75	18860	62.8
200	5.20	21600	78.8
200	17.53	21750	58.1

Table 2

Coefficients of the fit parameters $y = a + be^{-c \dot{\epsilon}}$ for two-dimensional compact tension specimens where y and $\dot{\epsilon}$ represent the normalized initiation or propagation fracture toughness and the strain rate, respectively.

2D Models	a	b	c	R ²
Normalized K _{init}	0.3636	0.4696	0.2854	0.80
Normalized R-curve Slope	0.1630	0.6570	2.772	0.78

Table 3

Coefficients of the fit parameters $y = a + be^{-c \dot{\epsilon}}$ for three-dimensional compact tension specimens where y and $\dot{\epsilon}$ represent the normalized initiation or propagation fracture toughness and the strain rate, respectively.

3D MicroCT Models	a	b	c	R ²
Normalized K_{init} (19-year-old)	0.3672	0.5512	0.3722	0.85
Normalized K_{init} (81-year-old)	0.3606	0.5745	0.2868	0.84
Normalized K_{init}^* (81-year-old)	0.2300	0.3667	0.2864	0.84
Normalized R-curve Slope (19-year-old)	0.2793	0.6159	0.8493	0.85
Normalized R-curve Slope (81-year-old)	0.1927	0.7328	2.124	0.79
Normalized R-curve Slope [*] (81-year-old)	0.1168	0.4443	2.124	0.79

* 81-year-old fracture toughness values normalized with respect to the maximum 19-year-old fracture toughness value