

The True Toughness of Human Cortical Bone Measured with Realistically Short Cracks

K. J. Koester,^a J. W. Ager III,^a and R. O. Ritchie^{a,b,*}

^a *Materials Sciences Division, Lawrence Berkeley National Laboratory, Berkeley, CA 94720*

^b *Dept. of Materials Science and Engineering, University of California, Berkeley, CA 94720*

Bone is more difficult to *break* than to *split*. Although this is well known, and many studies exist on the behavior of long cracks in bone, there is a need for data on the orientation-dependent crack-growth resistance behavior of human cortical bone which accurately assesses its toughness at appropriate size-scales. Here we use *in-situ* mechanical testing in the scanning electron microscope and x-ray computed tomography to examine how physiologically-pertinent short (<600 μm) cracks propagate in both the transverse and longitudinal orientations in cortical bone, using both crack-deflection/twist mechanics and nonlinear-elastic fracture mechanics to determine crack-resistance curves. We find that after only 500 μm of cracking, the driving force for crack propagation was more than *five times higher* in the transverse (breaking) direction than in the longitudinal (splitting) direction due to major crack deflections/twists principally at cement sheathes. Indeed, our results show that the true transverse toughness of cortical bone is far higher than previously reported. However, the toughness in the longitudinal orientation, where cracks tend to follow the cement lines, is quite low at these small crack sizes; it is only when cracks become several millimeters in length that bridging mechanisms can develop leading to the (larger-crack) toughnesses generally quoted for bone.

Bone is a complex hierarchical composite of collagen and hydroxyapatite (HA) that is imbued with mechanisms to resist fracture at different size-scales.¹ These size-scales relate to the characteristic structural dimensions in bone, which vary from twisted peptide chains at the nanoscale, HA-impregnated twisted collagen fibrils at the scale of tens of nanometers, collagen fibers that are typically a micron in diameter, the lamellar structure of collagen fibers at micron dimensions, to the (secondary) osteon (Haversian) structures

*Corresponding author: tel: (510) 486-5798; fax: (510) 643-5792. *E-mail address*: RORitchie@lbl.gov (R. O. Ritchie)

which are several hundred microns in size. It is the simultaneous operation of toughening mechanisms at these various length-scales that provide bone with its enduring strength and toughness. However, as the mechanical properties of bone can undergo deleterious changes with aging^{2,3} and disease,⁴ there is an imperative to understand, both quantitatively and mechanistically, the origins of its fracture resistance in order to develop therapies to inhibit or reverse these negative effects.

There are complications with this approach though when applied to human bone. Cortical bone develops its toughness primarily from extrinsic mechanisms^a during crack growth, as opposed to crack initiation;^{3,6,7} moreover, as its fracture behavior is intimately coupled to its hierarchical structure, the measured toughness will be a function of the size-scale at which it is assessed. In addition, measurements need to reflect cracking behavior in the more clinically-relevant transverse orientation, *i.e.*, to break rather than split bone, and involve realistic flaw sizes pertinent to human bones. Accordingly, in this work we have attempted to measure, for the first time, crack-resistance curves (R-curves) for human cortical bone in the transverse orientation, using *in-situ* measurements/observations in environmental scanning electron microscope (ESEM), coupled with fractographic and synchrotron x-ray computed tomography studies to identify the salient damage and toughening mechanisms. Our approach is three-fold: we utilize nonlinear-elastic fracture mechanics (NLEFM), with crack-deflection mechanics, to properly quantify the toughness properties, we have examined both crack-initiation and crack-growth (R-curve) toughnesses for realistically short crack sizes, and we have compared behavior to the more commonly evaluated longitudinal orientation.

There have of course been numerous previous evaluations of the toughness of mammalian cortical bone (see Table 1),⁸⁻²⁰ although none have focused on evaluating the important crack-growth properties for the short crack sizes associated with the cortical shell in the transverse orientation. Indeed, most prior measurements have involved single-

^a Fracture can be thought of as a mutual competition between *intrinsic* damage mechanisms, which operate ahead of the crack tip to promote cracking, and *extrinsic* toughening mechanisms, which act primarily in the crack wake to “shield” the crack from the applied driving force to inhibit cracking. Extrinsic toughening, which is most relevant for brittle materials (including bone), invariably results in resistance-curve behavior, *i.e.*, the need for a progressively increasing applied driving force to continue crack extension.⁵

parameter characterizations of the toughness,⁸⁻²⁰ e.g., K_c , which do not necessarily include any contribution from the crack-growth toughness; moreover, such measurements have, with few exceptions,^{2,19,20} been based on linear-elastic fracture mechanics,⁸⁻¹⁹ which minimizes any contribution from plastic (inelastic) deformation, or have involved large crack sizes (~several mm) in the longitudinal orientations (where it is easier to make measurements). There have been crack-growth resistance measurements^{3,7,21} (although none in human cortical bone in the transverse orientation) which have identified several microscale toughening mechanisms, including viscoplastic flow,²² microcracking,^{6,23,24} crack deflection,²⁵⁻²⁷ and crack bridging.^{21,28,29} Despite this body of research, there are still critical issues that remain to be assessed, namely that of the true crack-growth resistance of human cortical bone at physiologically-relevant short crack sizes, how this specifically changes with orientation, and a mechanistic understanding of the microstructural damage and toughening mechanisms involved.

Here, we use NLEFM testing of human cortical bone under hydrated conditions *in-situ* in the ESEM to permit resistance-curve measurements for growing short cracks, less than ~1000 μm in size, in both the transverse and longitudinal (proximal-distal) orientations, with simultaneous high-resolution imaging of crack paths to discern the dominant sources of toughness and to show how they actually confer crack-growth resistance. Our focus is on the transverse orientation because this is the more clinically relevant direction,^{1,30} however, cracks growing in this orientation invariably deflect along the longitudinal direction such that both orientations act in concert to provide resistance to fracture.

Using test samples from the midsection of the frozen cadaveric humeral cortical bone of three 37-41 year-old donors, both single-edged notched bend and compact-tension samples were sectioned and notched from locations longitudinal or transverse to the bone long axis. The notch orientation was such that the nominal crack-growth direction was either along the proximal-distal direction of the humerus, in the longitudinal-radial plane, *i.e.*, parallel to the long axis of the osteons (*longitudinal*), or transverse to the long axis of the humerus (*transverse*), as in Fig. 1. We monitored stable crack extensions ($\Delta a < 1000 \mu\text{m}$) in three-point bend samples soaked in Hanks' Balanced Salt Solution (HBSS), *in-situ* in the ESEM, and additional bend and C(T) samples *ex-situ* (outside the ESEM) in HBSS,

all at 25°C, to obtain information for larger crack extensions ($\Delta a \sim 150\text{-}7000 \mu\text{m}$); results are presented in terms of NLEFM measurements of the crack-driving force, specifically the stress-intensity factor K and J -integral (see Methods).^b

Our results showing the increase in crack-driving force as a function of crack extension, *i.e.*, the $K(\Delta a)$ resistance curves, for human cortical bone, are shown in Fig. 1, with simultaneous images of the crack profiles in Fig. 2. This is the first time that nonlinear-elastic fracture mechanics has been used to analyze the crack-resistance curve behavior of physiologically relevant short cracks in human bone. The degree of toughening in the transverse orientation is very large concomitant with gross crack deflection/twisting/meandering/delamination along the longitudinal and circumferential directions which results in extremely rough fracture surfaces (Fig. 2a-e). Stress intensities for crack extension reach values of $25 \text{ MPa}\sqrt{\text{m}}$ ($J \sim 31 \text{ kJ/m}^2$) over the first $500 \mu\text{m}$ of cracking, representing toughness values much higher than have previously been reported (Table 1). In contrast, corresponding fracture in the longitudinal orientation requires far lower driving forces, specifically, stress intensities between 1 and $2 \text{ MPa}\sqrt{\text{m}}$ ($J \sim 0.05\text{-}0.2 \text{ kJ/m}^2$) to initially extend the crack the first $\sim 500 \mu\text{m}$, consistent with the highly linear crack profiles and smooth fracture surfaces (Fig. 2f-j); however, with continued crack extension into the millimeter range, toughness values approach $5 \text{ MPa}\sqrt{\text{m}}$ ($J \sim 1.2 \text{ kJ/m}^2$), consistent with K_c values measured in most previous studies (see Table 1). We note here that the short-crack K values of ~ 1 to $2 \text{ MPa}\sqrt{\text{m}}$ for initial longitudinal crack propagation, although seemingly quite low, are consistent with a recent study of short-crack R-curve behavior, by Mullins *et al.*,³² who report values of 0.5 to $2.3 \text{ MPa}\sqrt{\text{m}}$ for 5 to $60 \mu\text{m}$ cracks in this orientation emanating from hardness indentations in ovine bone. We further note that our R-curve/toughness values in Fig. 1 are relatively insensitive to (i) changes in displacement rate, which was varied from 0.04 to $6 \mu\text{m/sec}$, (ii) any differences in hydration of our bone samples caused by the *in-situ* loading of HBSS-saturated specimens

^b The fracture toughness is generally defined in terms of a critical value of the stress intensity, K , which is a field parameter characterizing the distribution of local elastic stresses and displacements at a crack tip. It can be defined for three crack-displacement modes: mode I – tensile opening, mode II – shear, and mode III – anti-plane shear. Under nonlinear elastic conditions, *i.e.*, in the presence of some degree of inelasticity (*e.g.*, plastic deformation), the fracture toughness is better described in terms of the J -integral, which characterizes the corresponding distribution of local inelastic stresses and displacements at the crack tip³¹.

within the ESEM *vs.* the *ex-situ* loading of specimens immersed in HBSS, and (iii) any loss of constraint due to the small size of our *in-situ* bend samples (additional tests with a 7.5 mm thick specimen gave a similar transverse R-curve to the bulk of our data which were determined with 1.5 mm thick specimens).

The specific damage and toughening mechanisms associated with this behavior were imaged in real time through the use of *in situ* electron microscopy. We find that crack trajectories are highly deflected, both in-plane and out-of-plane (crack twist), in the transverse orientation and relatively linear in the longitudinal direction (Figs. 2-3). Specifically, for the transverse direction, cracks tend in general to deflect away from the nominal mode-I plane^c and then periodically to “delaminate” along internal interfaces, involving deflections of $\sim 90^\circ$ for tens to hundreds of micrometers, in the longitudinal direction, before reinitiating at a much higher applied load and continuing their deflected path along the (nominal) transverse direction; this process is repeated several times throughout the cortical bone specimen. In contrast, crack paths in the longitudinal direction remain comparatively linear along the nominal mode-I plane, although multiple (overlapping) cracks are generated with (microscopic) “uncracked ligaments”, some tens of micrometers in size, in between. Such uncracked regions are the source of crack bridging in bone, but are only fully developed as crack extension proceeds into the millimeter range.

Of particular note for these differing fracture paths is that surrounding the major cracks, there is clear evidence of microcracking, *i.e.*, smaller cracks typically tens of micrometers in length, within the bone matrix.^{33,34} These microcracks are primarily orientated along the longitudinal direction; they are therefore roughly parallel to the primary crack path in longitudinal specimens and perpendicular in transverse specimens (Fig. 2). Such microcracks form at “weak links” in the bone-matrix structure; they occur primarily along cement sheathes, *i.e.*, at the refractile boundaries of the Haversian systems (secondary osteons), and to a lesser extent at lamellar interfaces,³⁵ and correspondingly

^c The mode-I plane represents the plane of maximum tensile (tangential) stress, governed by a $K_{II} = 0$ path. For a perfectly homogeneous continuum, it represents the expected direction of cracking in a brittle material.

have a typical spacing in the tens to hundreds of micrometers aligned along the long axis of the bone.

We find that the formation of these microcracks, *and specifically their orientation*, are essential to the orientation-dependent fracture toughness of cortical bone; indeed, their presence forms the basis of the contrasting toughening behavior in the two orientations. In the transverse orientation, the microcracks at the cement lines are roughly aligned perpendicular to the crack path, where they act as “delamination barriers” (via the Cook-Gordon mechanism of crack arrest at weak interfaces³⁶); this serves to blunt the crack, cause crack deflections of $\sim 90^\circ$, generate highly tortuous crack paths (Figs. 2a-c,e), extremely rough fracture surfaces (Fig. 2d), and correspondingly high toughness. The latter follows because of the reduced local stress field due to blunting and the need to reinitiate the crack following perpendicular delaminations; in addition, such gross crack-path deviations away from the plane of maximum tensile stress greatly diminish the local stress intensity at the crack tip, thereby necessitating higher applied loads to continue cracking.

Conversely, in the longitudinal orientation, the microcracks are aligned roughly parallel to the growing crack; their formation alongside and ahead of the crack tip are the precursor to the formation of so-called “uncracked-ligament” bridges, which toughen by resisting the opening of the crack.²⁸ However, we do not believe that the microcracks *per se* toughen the material, via a constrained microcracking mechanism,^d as calculations suggest that the contribution to the toughness of cortical bone from this mechanism is relatively small.³⁸ Their vital role is to induce both primary forms of toughening in bone, that of crack deflection and crack bridging.

These observations of contrasting nature of damage and toughening in the transverse and longitudinal orientation are summarized in Fig. 2e,j) and in the movies contained in the Supplementary Information. However, as they are based on *in-situ* ESEM observations, they reflect the surface cracking behavior. To discern what is happening below the surface, we used synchrotron x-ray computed tomography to image the

^d Constrained microcracking is a mechanism where an array of microcracks forms around the crack tip locally dilating the material; the bulk of the material constrains this dilation and results in a compressive stress field around the tip of the crack serving to hold the crack tip closed.³⁷

underlying structure of the bone and to determine how the cracks interact with this structure. Specifically, three-dimensional images of crack propagation in transverse orientation, together with two-dimensional slices showing the crack at different depths throughout the sample, show the major crack-deflections/twists as the crack encounters the osteons (Fig. 3). Although the frequency of these major deflections appears in two-dimensional slices to be in the tens of microns; it is clear from three-dimensional imaging that this is controlled not by the in-plane spacing of the Haversian systems (\sim hundreds of microns), but by the spacing of the Haversian systems at different depths through the sample. If the Haversian systems from the three-dimensional imaging are projected onto a single plane, their spacing coincides with the frequency of these major deflections. Also of significance is that these three-dimensional images show that such crack deflection involves twisting of the crack, in addition to the in-plane tilts.

The importance of the short-crack ($<500 \mu\text{m}$) toughness properties cannot be overstated. For example, since the mid-shaft cortical shell of the humerus is only $\sim 3\text{-}5 \text{ mm}$ thick), it is unlikely that data on stable crack growth on the order of millimeters would be very relevant to *in-vivo* fracture. Moreover, parallel studies on dentin have shown that dominant toughening mechanisms identified at large crack sizes (up to 6 mm) do not necessarily reflect the physiologically-relevant fracture behavior of human tooth dentin where crack sizes are so much smaller (in the tens to hundreds of microns).³⁹ For the present measurements on bone, crack-initiation toughnesses, *i.e.*, where $\Delta a \rightarrow 0$, are comparable for both orientations ($\sim 1 \text{ MPa}\sqrt{\text{m}}$). With further crack extension, the crack-growth resistance will begin to increase as the scale of the relevant toughening mechanisms is encountered. In the longitudinal orientation, the toughness of 1 to $2 \text{ MPa}\sqrt{\text{m}}$ in the short-crack ($<500 \mu\text{m}$) regime essentially reflects the toughness of the cement sheathes; further crack extension into the millimeters range is required before significant uncracked-ligament bridges can develop such that the toughness approaches more expected numbers $\sim 5 \text{ MPa}\sqrt{\text{m}}$. In the transverse orientation, the characteristic size-scale pertains to the spacing of the osteons, which controls the process of crack deflection and twist at the cement lines; cracks must only extend $\sim 100 \mu\text{m}$ or so to activate this mechanism to cause rising R-curve behavior.

This process of major crack deflections at cement sheathes is clearly the most potent toughening mechanism in cortical bone. As noted above, three-dimensional tomographic imaging (Fig. 3) reveals extensive twisting of the crack path, in addition to in-plane ($\sim 90^\circ$) deflections, which substantially enhances the degree of toughening. Quantitatively, this toughening can be estimated using crack-deflection mechanics. The *local* mode-I and -II linear-elastic stress-intensities, k_1 and k_2 , at the tip of a deflected crack, can be stated in terms of the applied stress-intensities, K_I and K_{II} , by:⁴⁰

$$k_1 = c_{11}(\alpha)K_I + c_{12}(\alpha)K_{II} \text{ and } k_2 = c_{21}(\alpha)K_I + c_{22}(\alpha)K_{II}, \quad (1)$$

where $c_{ij}(\alpha)$ are mathematical functions of the deflection angle α . From Eq. 1, it can be calculated that one simple in-plane crack deflection by $\alpha \sim 90^\circ$ can reduce the stress intensity at the crack tip by roughly a factor of two, thereby effectively doubling the fracture toughness.³⁸ However, if we now incorporate crack twisting, the effect is much larger. The local stress intensities that result from the *twisting* of a *deflected* crack can be calculated by using crack-twist mechanics, where the effective *local* mode-I and -III stress intensities at the tip of the twisted and deflected crack are given by:⁴¹

$$k_I^{tw} = c_{11}(\phi)k_1 + c_{12}(\phi)k_{II} \text{ and } k_{III}^{tw} = c_{31}(\phi)k_1 + c_{32}(\phi)k_{II}, \quad (2)$$

where $c_{ij}(\phi)$ are mathematical functions of the twist angle ϕ and k_I and k_{II} are given in Eq. 1. Any combination of a twist of 45° and a deflection of 45° can halve the crack-tip stress intensity; however, a twist of 45° and a deflection of 90° , which is typical of transverse crack propagation in bone (Fig. 3), decreases the stress intensity at the crack tip by a factor of six.^e Accordingly, it is clear that such combinations of twist and deflection can lead to significant toughening in human bone, by factors on the order of two to six, consistent with the very rapid increase in crack-growth resistance in the transverse orientation (Figs. 1,3).^f This is undoubtedly why bone is much more difficult to break than to split.

^e It is perhaps worth noting here that the J -integral method used in the present work to evaluate the toughness bone is ideally suited for characterizing such mixed-mode fracture. Unlike the K approach, the energies associated with each three modes (mode-I, -II, -III) of failure can simply be summed to give a NLEFM driving-force for mixed-mode fracture: $J = K_I^2/E' + K_{II}^2/E' + K_{III}^2/2G$, where E' and G are, respectively, the appropriate Young's and shear moduli.

^f The toughening mechanisms of twist and deflection in bone, which are accounted for in our approach, are why we have obtained toughnesses approximately double that of others who have used nonlinear-elastic fracture mechanics but with side-grooved samples;²⁰ the presence of side-grooves specifically acts to inhibit such crack twisting.

METHODS

SAMPLE PREPARATION

Test samples from the midsection of frozen cadaveric humeral cortical bone were wet sectioned using a low speed saw and machined into seventeen 1.5 mm thick, 8 mm long bend samples (width $W = 2$ mm) and five 1.2–3.3 mm thick compact-tension C(T) samples ($W = 13$ -18 mm). Samples were then taken from locations longitudinal or transverse to the bone long axis, and notched to form an initial crack of roughly half the sample width, which was then sharpened with a micronotching technique using a razor blade irrigated with 1- μ m diamond suspension. The orientation of the notch was such that the nominal crack-growth direction was either in along the proximal-distal direction in the longitudinal-radial plane (*longitudinal* orientation), or transverse to the long axis of the humerus (*transverse* orientation). All samples were wet polished with an increasingly higher finish to a final polish with 0.05 μ m diamond suspension before being immersed in ambient HBSS for 4-24 hr prior to testing.

FRACTURE TOUGHNESS/R-CURVE MEASUREMENTS

In-situ testing of samples soaked in HBSS was performed for stable crack extensions less than ~ 600 μ m for the longitudinal orientation and ~ 150 μ m in the transverse orientation in a Hitachi S-4300SE/N ESEM (Hitachi America, Pleasanton, CA) at 25°C using a Gatan Microtest 2kN three-point bending stage (Gatan, Abington, UK); images of the crack path were obtained simultaneously in backscattering mode at 15 kV and a pressure of 35 Pa. In addition, we tested samples *ex-situ* in HBSS at 25°C on an EnduraTec Elf 3200 testing machine (BOSE, Eden Prairie, MN) and an MTS 810 (MTS Corporation, Eden Prairie, MN) to determine the R-curve for larger crack extensions ($\Delta a \sim 150$ -7000 μ m).

Fracture toughness R-curves were determined in terms of the crack-driving force as a function of crack extension (Δa). Where crack paths remained relatively linear and did not undergo large deflections, *i.e.*, they extended approximately along the expected path of the maximum tensile stress, as for most samples tested in the longitudinal orientation, standard handbook linear-elastic mode-I solution for cracks in these geometries⁴² were used to obtain the stress intensity. For samples tested in the transverse orientation, conversely, cracks grew at an angle to the mode I plane (of maximum tensile stress). Such crack deflection induces mixed-load loading at the crack tip, *i.e.*, mode-I (tensile-opening) plus mode-II (shear), such that standard mode-I stress intensity solutions are not applicable. For the first 100-200 μ m of crack extension, the cracks grew away from the notch at a constant angle such that standard crack deflection solutions⁴⁰ could be used to

compute a mixed-mode driving force, as has been done in human dentin.³⁹ With additional crack extension in the transverse orientation, cracks often deflected several times. To determine the R-curve for these more complex crack paths, a nonlinear-elastic fracture mechanics approach was used in which the crack-driving force was calculated using the J -integral⁴² and crack lengths were estimated in terms of the equivalent through-thickness crack with the same compliance. (Typically, for a deflected crack with a total length of 850 μm , where the projected length along the nominal mode I plane would be on the order of 625 μm ; the computed length of the linear crack with equivalence compliance would in between these values, within $\sim 10\%$ of the total length). This approach accounts for the contribution from “plasticity” to the toughness and for mixed-mode loading conditions, and provides a sound means to determine the R-curve fracture-toughness in a material that undergoes multiple large-scale crack deflections. Specifically, to monitor crack extension, measurements of the elastic compliance, C_{LL} , were made during periodic unloading ($\sim 10\%$) every $\sim 25\mu\text{m}$ of crack extension during the R-curve test. The relationship between C_{LL} and the equivalent through-thickness crack length, a , was obtained from handbook solutions.⁴² The stress intensity at each measured crack length was calculated by measuring J . J , which is equivalent to G under elastic conditions, is defined as the sum of the elastic and plastic contributions:⁸

$$J = J_{el} + J_{pl}. \quad (1)$$

The plastic component of J was calculated from⁴²

$$J_{pl} = \frac{2A_{pl}}{Bb}, \quad (2)$$

where A_{pl} is the area under the plastic region of the load vs. load-point displacement curve, B is the specimen thickness, and b is the (macroscopic) uncracked ligament ($W - a$); the corresponding elastic component given by $J_{el} = K^2/E'$, was small and typically only 5-10% of J_{pl} . However, since it is unusual to express the toughness of biological materials such as bone in terms of J , equivalent (effective) stress intensities were then computed from the standard J - K equivalence (mode I) relationship:

$$K_J = \sqrt{JE}, \quad (3)$$

with the Young’s modulus for bone taken as $E = 20 \text{ GPa}$.

⁸ J , the nonlinear strain-energy release rate, can be defined as the rate of change in potential energy under nonlinear elastic conditions for unit increase in crack area.⁴³

We believe that the approach used in the current work is a necessary for a full understanding of the fracture properties of human cortical bone; because we have examined short cracks in both the longitudinal and transverse orientations, utilized a R-curve analysis to capture both crack *initiation* and *growth* behavior, accounted for the “plasticity” and mixed-mode loading using the *J*-integral, and simultaneously characterized the development of the extrinsic toughening mechanisms in cortical bone.

References:

1. Currey, J.D. *Bones*, (Princeton University Press, Princeton, 2002).
2. Zioupos, P. & Currey, J.D. Changes in the stiffness, strength, and toughness of human cortical bone with age. *Bone* **22**, 57-66 (1998).
3. Nalla, R.K., Kruzic, J.J., Kinney, J.H. & Ritchie, R.O. Effect of aging on the toughness of human cortical bone: Evaluation by R-curves. *Bone* **35**, 1240-1246 (2004).
4. Giannoudis, P., Tzioupis, C., Almalki, T. & Buckley, R. Fracture healing in osteoporotic fractures: Is it really different? A basic science perspective. *Injury scientific basis of fracture healing: An update* **38**, S90-S99 (2007).
5. Ritchie, R.O., Gilbert, C.J. & McNaney, J.M. Mechanics and mechanisms of fatigue damage and crack growth in advanced materials. *Int. J. Solids Struct.* **37**, 311-329 (2000).
6. Vashishth, D., Tanner, K.E. & Bonfield, W. Experimental validation of a microcracking-based toughening mechanism for cortical bone. *J. Biomech.* **36**, 121-124 (2003).
7. Vashishth, D. Rising crack-growth-resistance behavior in cortical bone: Implications for toughness measurements. *J. Biomech.* **37**, 943-946 (2004).
8. Melvin, J.W. & Evans, F.G. Crack propagation in bone. *Biomechanics Symp. ASME* 87-88 (New York, 1973).
9. Bonfield, W. & Datta, P.K. Fracture toughness of compact bone. *J. Biomech.* **9**, 131-134 (1976).
10. Wright, T.M. & Hayes, W.C. Fracture mechanics parameters for compact bone--effects of density and specimen thickness. *J. Biomech.* **10**, 419-425 (1977).
11. Bonfield, W., Grynbas, M.D. & Young, R.J. Crack velocity and the fracture of bone. *J. Biomech.* **11**, 473-479 (1978).
12. Behiri, J.C. & Bonfield, W. Fracture mechanics of bone--the effects of density, specimen thickness and crack velocity on longitudinal fracture. *J. Biomech.* **17**, 25-34 (1984).
13. Moyle, D.D. & Gavens, A.J. Fracture properties of bovine tibial bone. *J. Biomech.* **19**, 919-927 (1986).
14. Norman, T.L., Vashishth, D. & Burr, D.B. Mode I fracture toughness of human bone. in *Advances in Bioengineering*, Vol. 20 (ed. Vanerby) 361-364 (ASME, New York, 1991).
15. Norman, T.L., Vashishth, D. & Burr, D.B. Fracture toughness of human bone under tension. *J. Biomech.* **28**, 309-320 (1995).
16. De Santis, R. *et al.* Bone fracture analysis on the short rod chevron-notch specimens using the x-ray computer micro-tomography. *J. Mater. Sci. - Mater. M* **11**, 629-636 (2000).

17. Phelps, J.B., Hubbard, G.B., Wang, X. & Agrawal, C.M. Microstructural heterogeneity and the fracture toughness of bone. *J. Biomed. Mater. Res.* **51**, 735-741 (2000).
18. Wang, X., Shen, X., Li, X. & Mauli Agrawal, C. Age-related changes in the collagen network and toughness of bone. *Bone* **31**, 1-7 (2002).
19. Malik, C.L., Stover, S.M., Martin, R.B. & Gibeling, J.C. Equine cortical bone exhibits rising R-curve fracture mechanics. *J. Biomech.* **36**, 191-198 (2003).
20. Yan, J., Mecholsky, J., John J. & Clifton, K.B. How tough is bone? Application of elastic-plastic fracture mechanics to bone. *Bone* **40**, 479-484 (2007).
21. Nalla, R.K., Kruzic, J.J., Kinney, J.H. & Ritchie, R.O. Mechanistic aspects of fracture and R-curve behavior in human cortical bone. *Biomaterials* **26**, 217-231 (2005).
22. Fantner, G. *et al.* Sacrificial bonds and hidden length dissipate energy as mineralized fibrils separate during bone fracture. *Nature Mater.* **4**, 612-616 (2005).
23. Vashishth, D., Behiri, J.C. & Bonfield, W. Crack growth resistance in cortical bone: Concept of microcrack toughening. *J. Biomech.* **30**, 763-769 (1997).
24. Vashishth, D. Rising crack-growth-resistance behavior in cortical bone: Implications for toughness measurements. *J. Biomech.* **37**, 943-946 (2004).
25. Brown, C.U., Yeni, Y.N. & Norman, T.L. Fracture toughness is dependent on bone location - a study of the femoral neck, femoral shaft, and the tibial shaft. *J. Biomed. Mater. Res.* **49**, 380-389 (2000).
26. Parasamian, G. & Norman, T. Diffuse damage accumulation in the fracture process zone of human cortical bone specimens and its influence on fracture toughness. *J. Mater. Sci. - Mater. M* **12**, 779-783 (2001).
27. Peterlik, H., Roschger, P., Klaushoffer, K. & Fratzl, P. From brittle to ductile fracture of bone. *Nature Mater.* **5**, 52-55 (2006).
28. Nalla, R.K., Kinney, J.H. & Ritchie, R.O. Mechanistic fracture criteria for the failure of human cortical bone. *Nature Mater.* **2**, 164-168 (2003).
29. Yeni, Y.N. & Fyhrie, D.P. A rate-dependent microcrack-bridging model that can explain the strain rate dependency of cortical bone apparent yield strength. *J. Biomech.* **36**, 1343-1353 (2003).
30. Behiri, J.C. & Bonfield, W. Orientation dependence of the fracture mechanics of cortical bone. *J. Biomech.* **22**, 863-867 (1989).
31. Knott, J.F. *Fundamentals of Fracture Mechanics*, (Butterworth & Co., London, 1976).
32. Mullins, L., Bruzzi, M. & McHugh, P. Measurement of microstructural fracture toughness of cortical bone using indentation toughness. *J. Biomech.* **40**, 3285-3288 (2007).
33. Burr, D.B. & Martin, R.B. Calculating the probability that microcracks initiate resorption spaces. *Journal of Biomechanics* **26**, 613-616 (1993).
34. Taylor, D., Hazenberg, J.G. & Lee, T.C. Living with cracks: Damage and repair in human bone. *Nature Materials* **6**, 249-317 (2007).

35. Yeni, Y. & Norman, T.L. Calculation of porosity and osteonal cement line effects on the effective fracture toughness of cortical bone in longitudinal crack growth. *J Biomed Mater Res* **51**, 504-509 (2000).
36. Cook, J., Gordon, J.E., Evans, C.C. & Marsh, D.M. A mechanism for the control of crack propagation in all-brittle systems. *Proceedings of the Royal Society of London. Series A, Mathematical and Physical Sciences* **282**, 508-520 (1964).
37. Evans, A.G. & Faber, K.T. Crack-growth resistance of microcracking brittle materials. *J. Am. Ceram. Soc.* **67**, 255-260 (1984).
38. Nalla, R.K., Stölken, J.S., Kinney, J.H. & Ritchie, R.O. Fracture in human cortical bone: Local fracture criteria and toughening mechanisms. *J. Biomech.* **38**, 1517-1525 (2005).
39. Koester, K.J., Ager III, J.W. & Ritchie, R.O. The effect of aging on crack-growth resistance and toughening mechanisms in human dentin. *Biomaterials* **29**, 1318-1328 (2008).
40. Cotterell, B. & Rice, J.R. Slightly curved or kinked cracks. *Int. J. Fracture* **16**, 155-169 (1980).
41. Faber, K.T. & Evans, A.G. Crack deflection process - I theory. *Acta Metall.* **31**, 565-576 (1983).
42. E1820. *Standard test method for measurement of fracture toughness*, (American Society for Testing and Materials, West Conshohocken, PA, 2006).
43. Rice, J.R. A path independent integral and the approximate analysis of strain concentration by notches and cracks. *J. Appl. Mech.* **35**, 379-386 (1968).
44. Yang, Q.D., Cox, B.N., Nalla, R.K. & Ritchie, R.O. Re-evaluating the toughness of human cortical bone. *Bone* **38**, 878-887 (2006).

Acknowledgments

This work was supported by the Director, Office of Science, Office of Basic Energy Sciences, Division of Materials Sciences and Engineering, of the U.S. Department of Energy under contract no. DE-AC02-05CH11231, and by the Laboratory Directed Research and Development Program of Lawrence Berkeley National Laboratory (LBNL). Computed x-ray tomography was performed at LBNL's Advanced Light Source, which is supported under the same contract. The authors thank Holly Barth for help in preparing the x-ray computed tomographs.

Competing interests statement

The authors declare that they have no competing financial interests.

Table 1. Single-value toughness measurements of mammalian cortical bone in the longitudinal and transverse orientations.⁸⁻²⁰

Bone	Orientation	K_c (MPa \sqrt{m})	Geometry	Authors	Year
Bovine Femur	Long	3.21	SENT	Melvin and Evans	1973
Bovine Femur	Trans	5.6	SENT	Melvin and Evans	1973
Bovine Femur	Trans	2.2-4.6	SENT	Bonfield and Datta	1976
Bovine Femur	Long	3.62	C(T)	Wright and Hayes	1977
Bovine Femur	Long	2.4-5.2	C(T)	Bonfield et al.	1978
Bovine Tibia	Long	2.8	C(T)	Behiri and Bonfield	1984
Human Tibia	Long	2.4-5.3	C(T)	Bonfield et al.	1984
Bovine Tibia	Trans	11.2	SENB	Moyle and Gavens	1986
Human Tibia	Long	3.7	C(T)	Norman et al.	1991
Bovine Tibia	Long	7.2	C(T)	Norman et al.	1991
Bovine Tibia	Long	6.2-6.7	C(T)	Norman et al.	1995
Human Femur	Trans	6.4	SENB	Zioupos et al.	1997
Bovine Femur	Long	4.9	CNT	De Santis	2000
Baboon Femur	Long	1.8	C(T)	Phelps et al.	2000
Baboon Femur	Trans	6.2	SENB	Phelps et al.	2000
Human Femur	Long	5.1	SENB	Wang et al.	2002
Equine Metatarsal	Trans	6.0-9.0	C(T) ^g	Malik et al.	2003
Bovine Femur	Long	2.6	C(T) ^g	Yan et al.	2007
Bovine Femur	Trans	5.1	C(T) ^g	Yan et al.	2007
Bovine Femur	Long	6.2*	C(T) ^g	Yan et al.	2007
Bovine Femur	Trans	10.5*	C(T) ^g	Yan et al.	2007

SENT, single edged notched tension; SENB, single edged notched bend; C(T), compact tension
*denotes a value of K_c back-calculated from J_c .

^gdenotes side-grooved specimens

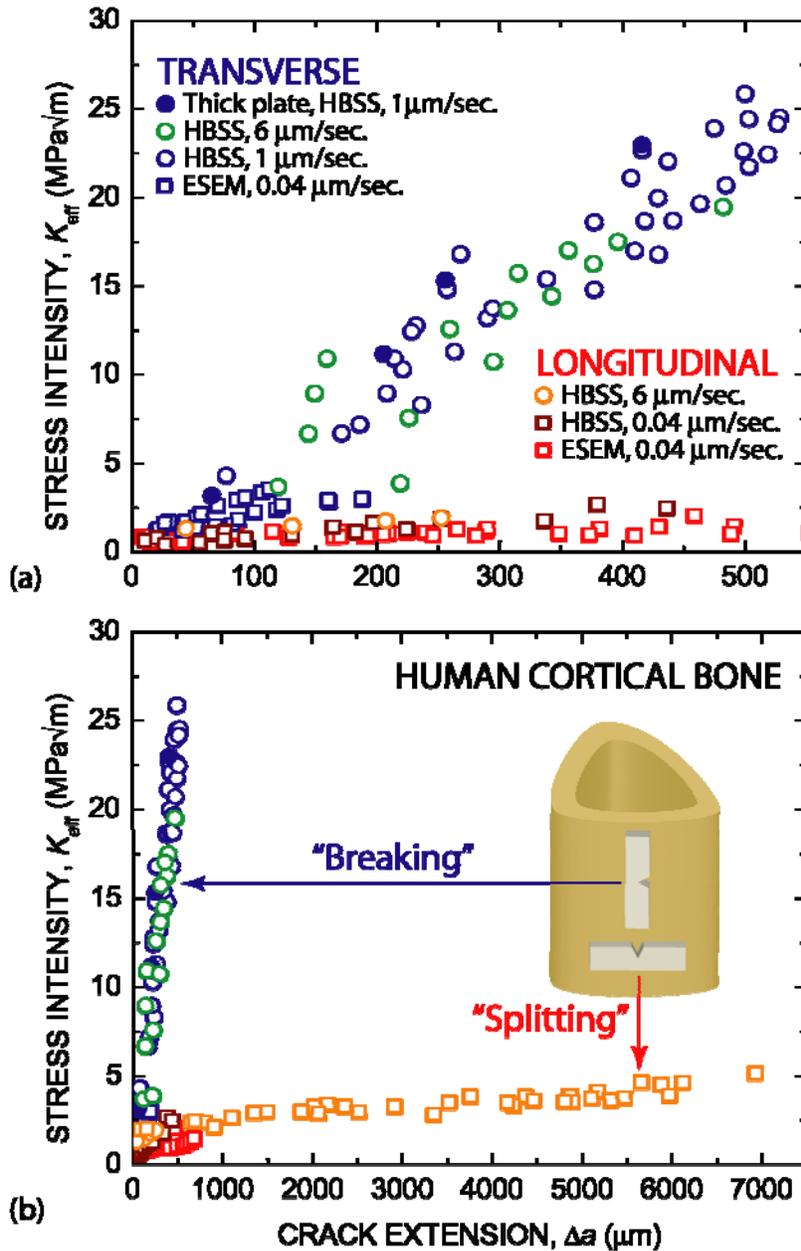


Figure 1. R-curves for the transverse and longitudinal orientations in human cortical bone. In the legend, ESEM denotes tested *in-situ* in the ESEM and HBSS denotes testing *ex-situ* immersed in HBSS; individual displacement rates are also given. In (a), are the R-curves for crack lengths $< 550 \mu\text{m}$ and in (b) are the combined short and long crack R-curves. The inset shows the orientation of the samples from the humerus. It is clear from the plots that the resistance to crack propagation increases much more rapidly in the transverse direction than in longitudinal direction. This was found to be the result of different toughening mechanisms in each orientation; deflection and twist in the transverse orientation and bridging in the longitudinal orientation. It was verified that this was not an effect of hydration, displacement rate, or loss of constraint by measuring R-curves *in-situ* and *ex-situ*, at different displacement rates, and a thick plate sample, respectively. As can be seen from (a) and (b), these factors did not affect the R-curves in either orientation.

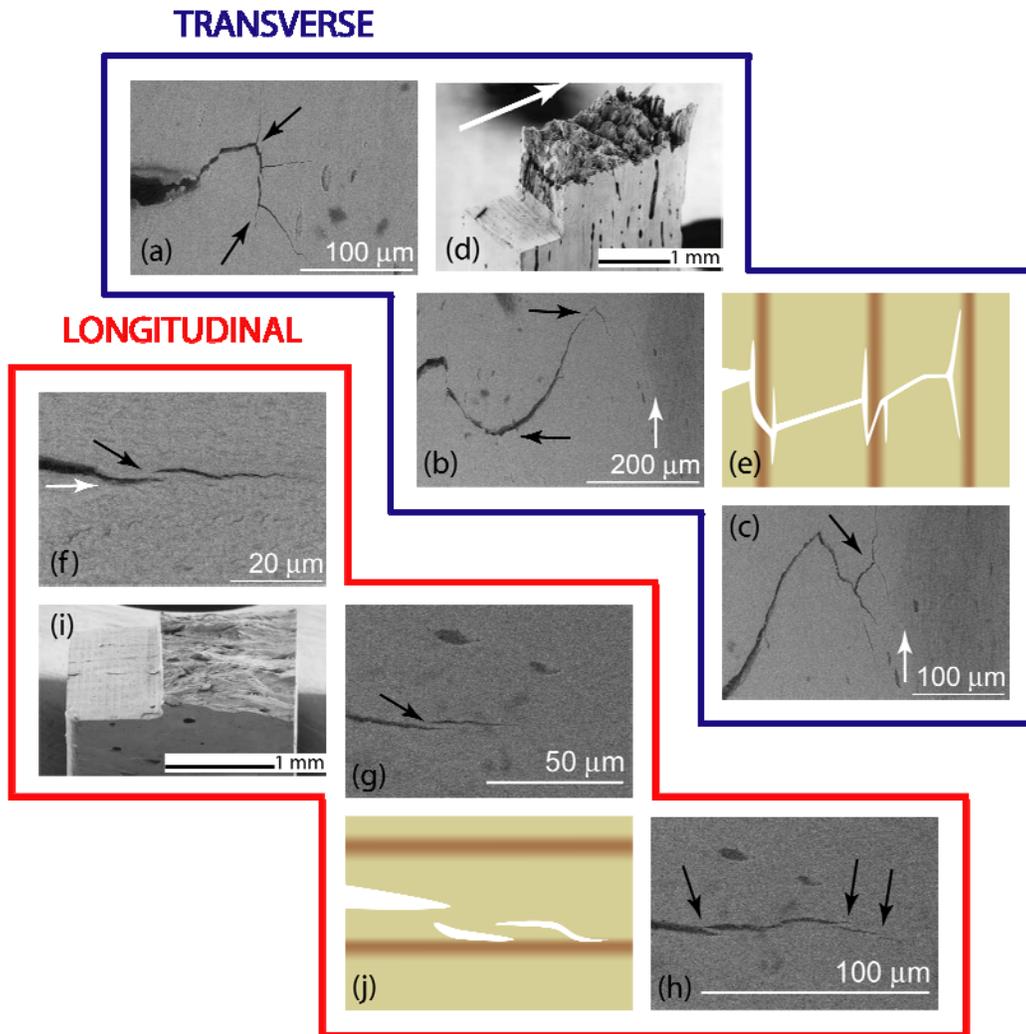


Figure 2. Crack profiles for representative cracks, at three different crack lengths, in the transverse and longitudinal directions accompanied by an ESEM fractography image and a schematic of the crack trajectory for each orientation. In images (a-c), the black arrows indicate the major in-plane deflections of the crack and the white arrows indicate the cement sheath. It can be seen from (a-c) that cracks in the transverse direction undergo multiple in-plane deflections, which in conjunction with the through-thickness twists (see Fig. 3 and the end of the transverse video in the Supplementary Information), give rise to the rough fracture surface in (d). A schematic of the crack profile is shown in (e), where the crack deflects at the cement sheaths as it propagates through the Haversian bone. Images (f-h) the black arrows indicate the formation of uncracked-ligament bridges, and the white arrow in (f) indicates the cement sheath. In contrast to the transverse direction, in the longitudinal direction the cracks do not deflect at the cement sheaths but follow them. At the cement sheaths in this orientation it is common for a new crack to initiate ahead of the propagating crack, resulting in the so-called “mother-daughter” crack configuration⁴⁴ and the formation of bridges. The amount and size of these bridges was found to be crack-size dependent; the larger the crack, the greater the extent of bridging. The cracks following the cement sheaths gives rise to the relatively flat fracture surface in (i). This process is shown schematically in (j), where the crack follows the cement sheaths and initiates new cracks ahead of the crack-tip leading to bridging.

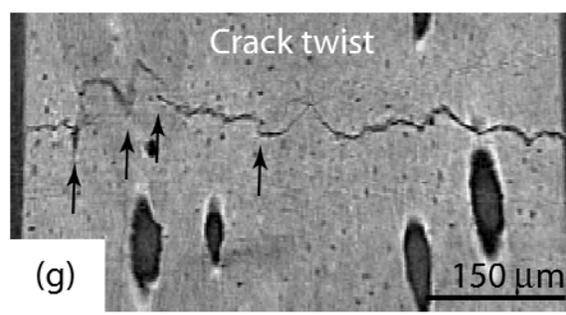
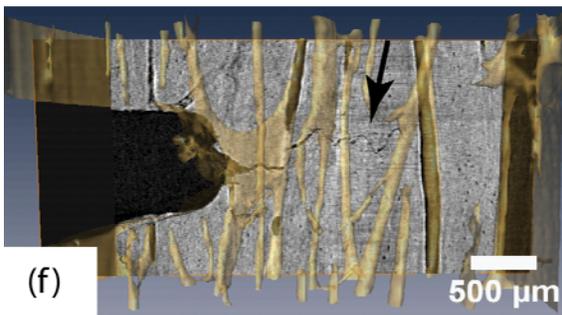
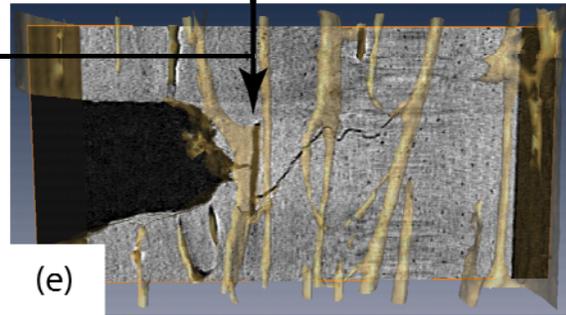
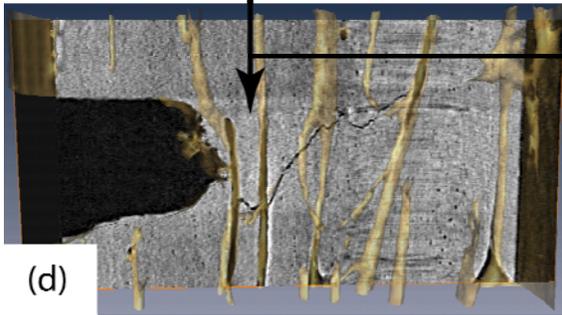
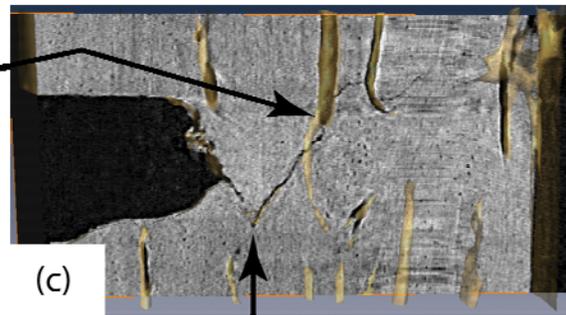
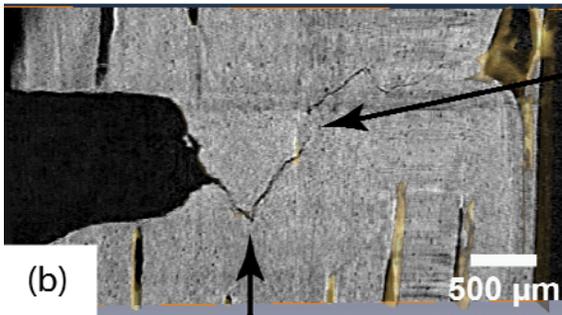
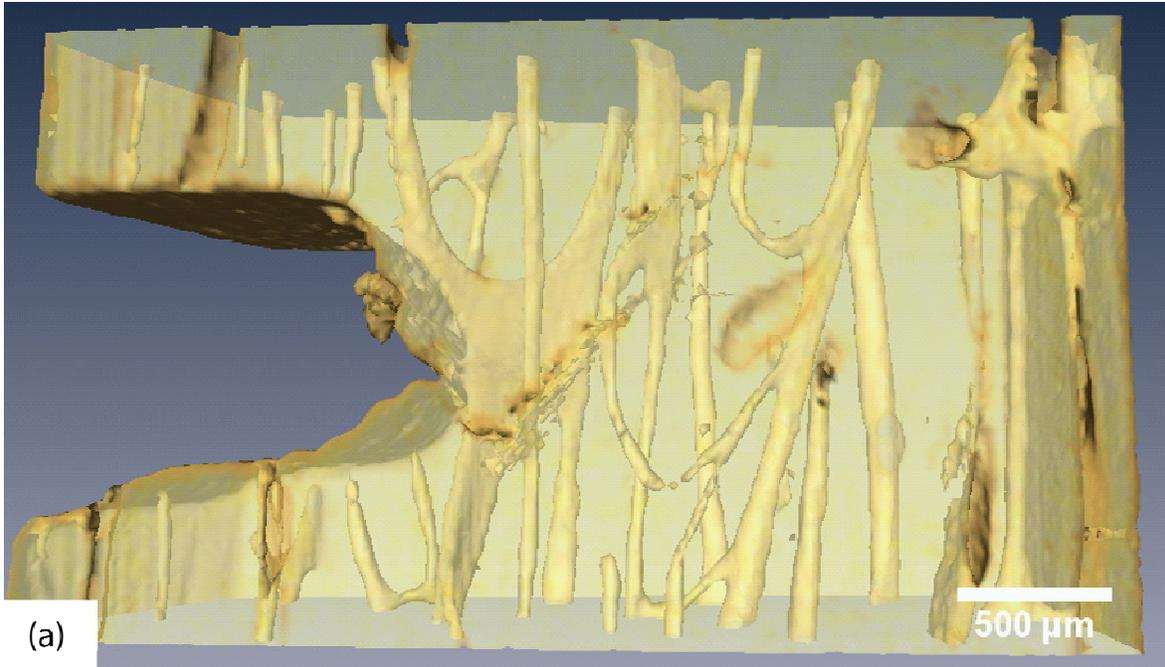


Figure 3. Synchrotron x-ray computed tomography images of the profiles of a crack growing in the transverse orientation, showing the process of crack deflection and twisting as the crack encounters the Haversian systems. Image (a) is a three-dimensional reconstruction of the sample in which the Haversian canals are partially transparent. Images (b-f) show a series of tomographic slices through the thickness of the sample combined with the transparent three-dimensional reconstruction of the sample, starting with the front face in (b) and moving to the back face (f); at depths of (b) 289 μm , (c) 342 μm , (d) 387 μm , (e) 507 μm , and (f) 587 μm . In (b-f) the arrows indicate the in-plane deflections and the associated underlying Haversian systems. The arrow connecting the features in (b) and (c) show that the 90° deflection of the crack is related to a Haversian system at a greater depth. The initial downward deflection of the crack as it grew off of the notch is related to the three Haversian systems deeper in the sample. Interestingly, this deflection is even sharper in the plane of the Haversian canal, as shown in (e). In (f) it can be seen that the crack has a markedly different trajectory and undergoes multiple in-plane deflections for shorter distances, as indicated by the arrow. In (g) a through-thickness slice from the front face to the back face is shown near the crack tip to highlight the twist of the crack through the sample. The arrows in (g) indicate the major twists, $\sim 90^\circ$, it can be seen that in addition to the twists of this magnitude the crack twists through the entire thickness at lower angles. This combination of in-plane deflections and through-thickness twists are related to the Haversian structure of bone and give rise to the rough fracture surface in Fig. 2(d) and are dominant source of the high fracture resistances reported in Fig. 1.