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Contribution of the cortex to epiphyseal strength

The upper tibia studied in cadavers

Non-destructive measurements of compressive stiffness were carried out on 20 proximal tibial autopsy specimens. The tibial epiphyses were first loaded through a template covering all but the peripheral 2 mm of the subchondral resection surface, then through the whole resection surface, and finally, after removal of the peripheral shell. A slight increase of the stiffness coefficient resulted from peripheral contact. Stiffness increased significantly after removal of the shell, but several potential sources of systematic error in this part of the investigation raise questions as to the validity of this finding. The area-corrected stiffness showed a decrease as a result of peripheral contact; this result indicates that the peripheral rim of bone has a lower area-corrected stiffness than the central bone, a finding which is incompatible with the concept of a true cortical shell at the epiphyseal level.

Ivan Hvid
Jørn Jensen
Strange Nielsen

Biomechanics Laboratory,
Orthopaedic Hospital, Randersvej 1, DK – 8200
Århus N, Denmark

Although full coverage of the tibial resection surface is not a realistic option with most arthroplasty designs, some clinicians stress the importance of full area contact (Freeman et al. 1982). Experimental work indicates that the tibial component, whether all polyethylene or metal-backed, should preferably cover a large fraction of the tibial resection surface (Bargren et al. 1978, Reilly et al. 1982, Figgie et al. 1984). A substantial load-bearing function of the cortical shell at the epiphyseal level was suggested by Figgie et al. (1984). Analytical models of the proximal tibia sometimes include an epiphyseal cortical bone shell (Murase et al. 1983), while other authors believe that the cancellous bone alone supports the joint load at this level (Askew & Lewis 1981, Bartel et al. 1982, Lewis et al. 1982).

We have studied the mechanical significance of the peripheral rim of bone delineating the cross-section of the proximal tibial epiphysis.

Material and methods

Twenty-three knee joints were obtained from routine autopsies within 24 h postmortem and immediately deep-frozen to below −21°C. After thawing, the tibiae were isolated, and the joint surfaces were resected closely beneath the subchondral bone plate. A parallel cut produced approximately 25-mm thick specimens. The proximal resection surface was copied onto paper, and a secondary surface excluding a peripheral 2 mm rim was constructed. Each area was measured twice with a planimeter. From the secondary surfaces, 4-mm thick steel templates were manufactured.

The mechanical tests were run in a universal materials testing machine (Instron® 1195). The specimen was placed on a rigid metal block. The load cell was equipped with a heavy steel column, slightly curved at the end facing the specimen to ensure uniform axial loading. Between the specimen and the loading column, the 4-mm steel template or a 4-mm steel plate covering the whole resection surface was inserted. To disperse the load equally, a 19-mm steel disc was added on top of the template (Figure 1). Finally, the peripheral bone was removed carefully from the proximal 5 mm of the epiphysis with a sharp osteotome to investigate the possibility that the shell contributes to epiphyseal compressive stiffness by resisting lateral expansion. Each loading condition was tested twice with an approximately 45-min interval. The individual compression test was stopped when a linear graphic output had been obtained. Between tests the specimens were kept in physiological saline. The individual test was performed as one "conditioning" loading cycle followed by the measuring cycle; conditioning was necessary because the stiffness of bone is load-history dependent (Zilch et al. 1980). The increase in stiffness between the first and second load cycle amounted to as

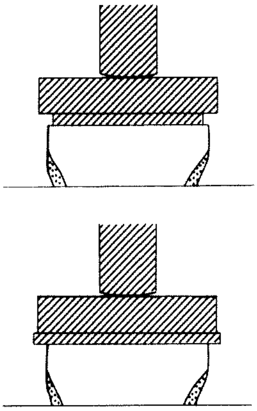

Figure 1. Diagram showing the application of load excluding the peripheral rim of bone (above) and including the peripheral rim (below). Steel plates and piston shown hatched, cortical bone shown dotted.

much as 50 per cent. The last load application in each series was carried beyond the ultimate failure point. Thus, it was possible to ensure that the non-destructive tests were within the linear range of the curve as assumed. This was the case with 20 specimens.

The deformation rate was kept constant at 0.5 cm min^{-1} (approximate strain rate 0.003 s^{-1}). Also, the recording paper was advanced at a constant speed. Thus, the time axis could be converted to deformation correcting for machine compliance.

From the force (F) and deformation (D) data, the stiffness (S) was calculated:

$$S = F / D.$$

The area-corrected stiffness (S_A) was calculated from:

$$S_A = F / (D \times A)$$

where A is the contact area. It may be seen that if an increase in stiffness is caused by an equal percentual increase in area, then the area-corrected stiffness is unchanged. If the stiffness increase is less than expected from the increase in area, then the area-corrected stiffness is reduced. Statistical comparison of the test series was by Student's paired *t*-test.

Results

Ninety-five per cent tolerance limits for the area measurements were ±0.15 cm^2 (approximately ±0.5 per cent) and ±4.9 MNm^{-1} for the stiffness measurements (approximately ±10 per cent).

The average gain in stiffness from the addition of peripheral contact was 1.3 MNm^{-1} (Table 1). When the peripheral bone was removed, an increase in stiffness resulted (average increase = 4.2 MNm^{-1}, standard error of difference = 1.5 MNm^{-1}, $t = 2.85$, $P = 0.01$).

The gain in stiffness from peripheral contact was smaller than expected from the gain in contact area as seen from the comparison of area-corrected stiffness (Table 1).

The stiffness coefficients measured after removal of the shell correlated closely with the maximal force measured during the final, destructive test ($r = 0.91$, $t = 9.43$, $P < 0.0001$).

Discussion

With the tolerance limits shown above, an average stiffness difference of approximately 5 per cent would be statistically significant at the 5 per cent level. Cortical bone tested in compression is roughly 30 times stiffer per unit area than cancellous bone (Currey 1970). This means that a 0.1 mm shell (a realistic minimal thickness value, according to Murray et al. 1984) with cortical bone properties would be expected to increase the stiffness of a 30 cm^2 epiphysis by some 20 per cent. Thus, the design of the study would seem to be adequate to detect the effect of a cortical shell.

The increase in the stiffness coefficient resulting from peripheral contact did not reach a statistically significant level; the increase was much smaller than expected from the increase in area, since the area-corrected stiffness decreased more than 10 per cent on average. The results after removal of the peripheral bone do not suggest a significant role of the shell in resisting lateral deformation; in fact, the stiff-

Table 1. Stiffness with and without peripheral contact

Peripheral contact	Stiffness (MNm^{-1})	Area-corrected stiffness (GNm^{-3})
Excluded (E): mean	34.5	11.3
Included (I): mean	35.8	12.8
Mean difference (I-E)	1.3	−1.5
St. error of difference	0.9	0.4
Significance	P > 0.10	P < 0.001

ness increased after removal of the peripheral bone, a finding which is difficult to explain. Impaction of bone caused by the osteotome is one possible explanation, but this may not be the only factor involved. Although some consideration was given to the time-dependent properties of bone, perhaps this aspect was not fully controlled in our study, the time-stiffness relation during repeated non-destructive loading of cancellous bone being unknown. A review of the results of the double-determinations of stiffness coefficients suggests a systematic trend to increased stiffness being recorded at the second of the two determinations; since the tests were performed in the sequence stated, such a trend does not affect the conclusion that peripheral contact produces little increase in stiffness, but it might have added to an impaction effect in the last part of the experiment. Finally, the fact that the last stiffness value was measured from a destructive test graph (offering a longer linear interval) might also have caused a systematic error towards higher stiffness values.

The "measuring" load-cycle invariably exhibited a steeper linear slope than the "conditioning" load-cycle, with a maximal increase in the stiffness coefficient of approximately 50 per cent. Unfortunately, no comparable data are available in the literature, but it is apparent that other investigators have encountered the same difficulty when subjecting composite specimens to non-destructive tests (Rockoff et al. 1969, Krause et al. 1976, Bourne et al. 1980). Carter & Hayes (1977), examining machined cylindrical cancellous bone specimens, found that hydraulic strengthening was apparent only when the strain rate was extremely high, e.g. 10.0 s^{-1} or several thousand times faster than in the present investigation. Thus the strain rate used seems appropriate to *avoid* hydraulic strengthening. Furthermore, if hydraulic strengthening were present, it would be expected to affect consecutive load-cycles, including the first cycle, equally.

The close correlation between stiffness and strength in the last part of the experiment, a correlation that is well-documented in tests on cancellous bone specimens (Goldstein et al. 1983, Hvid & Jensen 1984), suggests that the findings may be extrapolated to include the strength of the epiphyseal structure.

Figgie et al. (1984), in an experimental study comparing the ultimate load-bearing capacity of paired tibiae with standard or conforming (full surface covering) tibial components, found that the increase in load-bearing capacity was proportional to the area increase. For some reason the largest available standard component (72 mm) was not included in the study, and thus some non-utilized areas were probably unrealistically large. The authors speculated that factors other than area may come into play when the prosthesis reaches the cortical shell, but this was not supported by their findings. Reilly et al. (1982) measured surface strain at the proximal tibia, and compared the normal state to a variety of prosthesis configurations. Relatively high proximal strain readings were found with full area contact, while strain readings were subnormal with partial contact. This may suggest an abnormally low *in vivo* loading of the uncovered peripheral bone with standard implants, but disregards the effect of ligaments and tendons inserting in this area; it does not necessarily suggest a significant load-bearing function of the most proximal shell since weaker bone is subjected to higher strain per unit load.

Finite element analysis of the effect of increasing area coverage by the tibial component – in a model that included a proximal cortical shell (Murase et al. 1983) – did not suggest a stress-lowering effect of larger tibial components unless metal-backed. This result is in agreement with the experimental finding of Walker et al. (1984) that the polyethylene tibial component transmits relatively high stresses beneath the areas contacting the femoral component, i.e. at the centers of the condyles.

Murray et al. (1984) investigated the mechanical properties of the proximal metaphyseal shell in a more direct manner. From plate-bending tests on machined specimens, an average elastic modulus of approximately 1.5 GPa was derived; this figure is one order of magnitude lower than reported for cortical bone (Currey 1970). Furthermore, the failure crack-pattern of the proximal metaphyseal specimens suggested isotropy, indicating that this bone is unlikely to resist significant compressive forces. The present findings are well explained by the material properties quoted above.

Our findings are not compatible with the concept of a true cortical bone shell surrounding the proximal tibial epiphysis. The potential benefit of including the peripheral rim of bone in the contact area is likely to be negligible, since the increase in stiffness is less than 5 per cent.

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