A BIOMECHANICAL ANALYSIS OF THE USE OF CEMENT TO ANCHOR THE FEMORAL HEAD PROSTHESIS

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The making of a permanent bond between an inert surgical implant and a living bone, which will remain sound even though transmitting the forces produced by the weight of the body, is one of the most urgent subjects for research in modern orthopaedics. An attempt is made here to explain in simple mechanical terms the success which has attended the use of cement in making these bonds, clinical reports of which have been published elsewhere (Charnley 1960, Charnley and Kettlewell 1965).

Elasticity of bone—When the weight of the body is transmitted to the shaft of the femur the bone will shorten as a result of elastic deformation, and when the load is removed the femur will return to its original length. Figures for the modulus of elasticity of bone are available in the literature and range from 2.9×10^6 pounds per square inch* to 1.5×10^6 pounds per square inch (Smith and Walmsley 1959). If we take 2.2×10^6 pounds per square inch as an average figure for the modulus of elasticity of bone and assume that the shaft of a femur is 15 inches long, has an external diameter of 1.25 inches and an internal diameter of 0.5 inch, we can calculate the contraction in length under a load of 150 pounds in the following way:

$$\begin{aligned}
 \text{Cross-sectional area of bone} &= \frac{\pi}{4}(1.25^2 - 0.5^2) \\
 &= 1.026 \text{ square inches} \\
 \text{Axial load} &= 150 \text{ pounds} \\
 \text{Compressive stress} &= \frac{150}{1.026} = 146 \text{ pounds per square inch} \\
 \text{Modulus of elasticity} &= 2.2 \times 10^6 \text{ pounds per square inch} \\
 \text{Strain (i.e., contraction per unit length of femur)} &= \frac{146}{2.2 \times 10^6} = 66.3 \times 10^{-6} \\
 &= .000066\dagger
 \end{aligned}$$

Therefore a femur 15 inches long will contract .00099 inch.

In other words, and in round figures, the femur in the above example will shorten by about one-thousandth of an inch under a load of 150 pounds (Example 1).

If a structure is composed of materials possessing different moduli of elasticity there is a possibility that when exposed to a heavy load the different parts will contract by different amounts, so that slight movement will take place at the junction between the two pieces of dissimilar material. In the same way, a structure composed of different parts, but all of the same material, may demonstrate movement between the parts because of differences in elastic deformation, if the design is such that the different parts are not equally stressed when the load is applied.

This phenomenon of minute relative movement occurring repeatedly between the parts of a structure which is subjected to variations in loading is known as "fretting" and lies at

* The physical constants related to the strength of bone have been taken from the *Instructional Course Lectures, American Academy of Orthopaedic Surgery*, Vol. XI, p. 95, 1954, quoted by Bechtol, Ferguson and Laing, 1959.

† For the benefit of readers not fully conversant with strength of materials theory, it may be of interest to note that the figure for "strain" (.000066) has no units, because it is a ratio of the deformation to the whole length of the femur.

the root of a number of causes of mechanical failure in engineering practice. In this paper an attempt is made to show that fretting between a prosthesis and living bone, as a result of differences in the stressing of different parts, may very well be the fundamental cause of past failures to maintain a permanent bond in implant surgery between metal and living bone.

Elementary features of a femoral head prosthesis—The elements of a femoral head prosthesis are represented in the model illustrated, in which, for the sake of simplicity, the weight of the body is directed along the axis of the bone (Fig. 1). The stem of the prosthesis has been represented as a parallel cylinder in order to emphasise the fact that the stem of a conventional prosthesis has little or no capacity to transmit load to the bone, since the interface between the polished stem and the inner surface of the bone is in almost pure shear. The stem of such a prosthesis functions merely as a stabiliser to eliminate lateral “wobble” and it is through the collar of the prosthesis that the load is transmitted to the cut end of the stump of the femoral neck.

The effect of applying a load to the simplified arrangement illustrated in Figure 1 is for the shaft of the femur to be compressed. The stem of the prosthesis below the level of the collar will be completely unstressed, because the load passes directly from the head of the prosthesis, through the collar, to the bone. From the previous calculation we saw that under a load of 150 pounds every inch of the femur will shorten by $\cdot 000066$ inch. If the stem of the prosthesis is 5 inches long the corresponding length of femoral shaft will shorten by $\cdot 000330$ inch. But the metal stem of the prosthesis, not being under load, will remain the same length. There will therefore be a fretting movement of $\cdot 00033$ inch between the prosthesis and the endosteal surface of the femur at the level of the tip (diminishing in amount in the direction of the level of the collar).

FIG. 1
Simplified arrangement
for a femoral head
prosthesis.

It is generally agreed that when a person stands on one leg the compression forces generated in the neck of the femur by the contraction of the abductor muscles holding the pelvis level are over twice the deadweight of the body. In dynamic action there will be transient compressive stresses very much higher than twice the weight of the body. If we multiply the above figures by a factor of 3 to allow for the static and dynamic action of the muscles of the hip we see that the range of fretting movement at the level of the tip of a prosthesis 5 inches long would be about $\cdot 00099$ or approximately one-thousandth of an inch. If we consider that one micron (μ) is $\frac{1}{25,000}$ of an inch we see that this range of fretting corresponds to 25μ . This gives some

idea of the magnitude of the fretting movement in terms of histological measurements, since if we take a red blood corpuscle to measure 5μ it means that the tip of the prosthesis will be fretting over a range equal to five red blood cells laid side by side (Example 2).

In the development of femoral head prostheses there have been numerous attempts to improve “fixation” by increasing the length of the prosthetic stem to pass a long distance down the medullary canal of the femur. While this procedure undoubtedly will enhance the stability of the prosthesis against lateral angulation, the above calculations reveal that it will increase very considerably the amount of fretting between the tip of the prosthesis and the endosteal bone at that level. Thus for a prosthesis with a stem reaching to the level of the knee (say 15 inches long) the fretting would be $\cdot 003$ inch or equal to 15 red blood cells laid side by side (Example 3).

Surface area and transmission of load—In the conventional prosthesis, with the stem driven into the medullary canal, like a nail, until the collar of the prosthesis abuts against the cut

end of the femoral neck, only two areas are concerned in transmitting force to the bone of the femur. The principal area is that offered by the cut surface of the neck of the femur (which is not necessarily as large as the area of the collar of the prosthesis provided for this purpose). The other area lies on the lateral surface of the distal part of the stem of the prosthesis in contact with the endosteal surface of the femur. For all practical purposes the collar of the prosthesis transmits the whole weight of the body to the bone because the lateral surface of the polished metal stem is in shear as regards the vertical component of the forces acting on the prosthesis. The force transmitted by the lateral surface of the distal stem of the prosthesis merely neutralises the moment of forces tending to cause the prosthesis to turn in a varus direction. This concept of the prosthesis as a lever with a fulcrum at the calcar femorale is illustrated diagrammatically in Figure 2 and the distal part of the stem of the

FIG. 2
Figure 2—Femoral head prosthesis loaded vertically by body weight in the absence of abductor muscle contraction. Prosthesis is a lever over fulcrum f , the calcar femorale, and stress on bone is related to the dimension Z , which is related to the length of the neck of the femur. The force T neutralises the moment of the force F tending to rotate the prosthesis into varus. The roller signifies absence of load transmission in the direction of the force F . FIG. 3
Figure 3—Femoral head prosthesis loaded by the combined action of body weight and abductor muscle tone G acting in the direction of the resultant force R which is normal to the cut surface of the femoral neck. Length of femoral neck unimportant in this arrangement. Stem of prosthesis purely a stabiliser in this arrangement.

prosthesis has been represented as containing a roller to emphasise its inability to transmit axial load. This diagram shows also how the direction of the resultant compressive force in the neck of the femur, and the length of the femoral neck interact. If the resultant compressive force on the neck of the femur (the resultant of the weight of the body and the pull of the abductor muscles) lies perfectly in the direction of the neck of the femur, the length of the femoral neck (long or short) does not modify the load on the bone at any point (Fig. 3). If, however, the abductor mechanism is defective and the resultant takes a more vertical direction (Fig. 2) the length of the femoral neck will be very important, a long neck increasing the bone pressure on the calcar and a short neck reducing the pressure to the minimum (which is that caused by the deadweight of the body in the absence of muscular action).

Surface area and compressive stress in bone—The previous calculations based on the elementary diagram of Figure 1 need to be modified because the surface area of the bone used in this

diagram is that presented in a section through the mid-shaft of a femur in a strong young adult. The cut surface of the femoral neck in a senile patient presents a cross-sectional area of cortical bone very much less than the 1.025 square inches calculated from a tube with external diameter of 1.25 inches and internal diameter of 0.5 inch. When the areas of cortical bone in the cut surfaces of the femoral necks shown in Figure 4 are measured on squared paper it is evident that the surface area available in the femoral neck is only one-third that in the mid-shaft. In these two examples the area of the cut surface of the femoral neck can be taken approximately as 0.2 square inch. The body weights of the subjects from whom these specimens were obtained unfortunately are not known, but since they were senile patients it would probably be safe to take 120 pounds as an approximation. This means that the cut section of the cortical bone in contact with a prosthesis will be stressed at $\frac{120}{0.2} = 600$ pounds

per square inch which is very much more than the 146 pounds per square inch which was calculated in the first theoretical example for the cross-section of the mid-shaft of the femur. If we multiply this by a factor of 3 to allow for the action of the abductor muscles and transient dynamic forces, this raises the stress on the cut surface of the cortical bone in the femoral neck to 1,800 pounds per square inch (Example 4).

This loading of 1,800 pounds per square inch is approximately one-sixth of the ultimate strength of bone (International Course, American Academy of Orthopaedic Surgery 1954) which ranges from a minimum of about 6,000 pounds per square inch in transverse shear to a maximum of 24,000 pounds per square inch in compression. It is of interest to note, despite these high figures for the ultimate strength of bone, that Frankel (1960) found that most femoral necks taken from cadavera fractured under loads of about 800 kilograms (1,760 pounds) when stressed in a direction corresponding to the force resultant from the weight of the body and the contraction of the abductor muscles.

If we recalculate, using the figures from Example 4 for the actual conditions in the neck of the femur as exemplified in the two specimens of Figure 4, we find the following:

Compressive stress in bone	= 1,800 pounds per square inch
Modulus of elasticity	= 2.2×10^6 pounds per square inch
Strain (i.e., contraction of shaft per unit length)	= $\frac{1,800}{2.2 \times 10^6} = .00081$.

The strain of .00081 in this calculation, based on conditions in a senile person's neck of femur, is nearly ten times greater than the strain of .000066 in the dense bone at the mid-shaft of the femur in Example 2. This implies that fretting between the tip of a prosthesis and the endosteal surface of the bone in a senile patient would be nearly ten times greater than in a young person. The difference in actual fact is not as great as a factor of ten because the thinness of the cortical bone of the senile neck of femur, which is the factor responsible for raising the stress to 1,800 pounds per square inch is confined to the upper part of the femur, and it does not affect the whole 5 inches of length of bone corresponding to the stem of the prosthesis. Nevertheless, fretting at the tip of the prosthesis in the senile patient, for the reasons in this calculation, will be considerably greater than that in the calculation of Example 2. (Example 5.)

FIG. 4

Area of cross-section of cancellous bone in the cut surfaces of two femoral necks, upper figures, and the midshafts of the femurs, lower figures, in the senile patients (ages, 75 left, and 81 right).

Weight transmission by the stem of a prosthesis—The previous calculations have been based on the assumption that it is through the collar of the prosthesis that the whole of the load of the weight of the body, plus the forces resultant from the action of the gluteal muscles, is transmitted to the cut end of the femur. There is one special circumstance where this does not apply; that is when the tapering stem of the prosthesis fits the medial to lateral diameter of the medullary canal so that it is wedged between the two cortices. When this occurs the collar of the prosthesis does not make full contact with the cut end of the femoral neck and the weight of the body, augmented by the forces of the abductor muscles, must therefore be transmitted to the bone of the shaft of the femur through the medial and lateral surfaces of the stem of the prosthesis.

To analyse this situation let us consider the stem of the prosthesis as a polished steel wedge 5 inches long, tapering from a width of 1 inch at the base to 0.25 inch at the tip. In round figures this corresponds to a taper of 1 in 7, as can be shown by a geometrical drawing—

FIGS. 5 AND 6

Figure 5—Geometrical construction to show frontal dimensions, in inches, of an idealised prosthetic stem of Thompson type. Figure 6—Idealised prosthesis transmitting axial load to bone through a "fir-tree" type of anchorage. Y is length and X is width of the cylinder, of surface area πXY square inches, exposed to shear force of $\frac{F}{\pi XY}$ pounds per square inch.

that is, the stem would be 7 inches long if it tapered to a point instead of being truncated at a width of 0.25 inch (Fig. 5).

If we neglect frictional resistance between the polished steel surface and the moist bone the progressive penetration of the steel wedge into the medullary canal of the femur, under the weight of the body augmented by abductor muscle power, will be arrested when the resistance to elastic dilation of the bony canal balances the axial load. In the absence of frictional resistance the steel wedge will return to its original level when the axial load is removed, as a result of elastic recoil of the bony tube.

The force available to dilate the bony canal, measured in a direction normal to the surface of the wedge (which is for all intents and purposes transverse to the axis of the bone) will be approximately seven times the axial load. If we take the example of a small patient, weighing say 120 pounds, we must multiply this figure by a factor of at least 3 to allow for the action of the gluteal muscles when standing on one leg and part of the transient dynamic forces

in walking. The total transverse force available to dilate the femoral canal is thus $120 \times 3 \times 7 = 2,520$ pounds. This force is acting normal to surfaces presented by the medial and lateral borders of the stem of the prosthesis. If these surfaces are plane surfaces (which in many designs of prostheses they are not) we can calculate the area, assuming the wedge to be five-sixteenths ($\cdot 3126$) inch in thickness and assuming that the whole length of both medial and lateral surfaces are in firm contact with bone. This area will be approximately $5 \times 2 \times \cdot 313 = 3.13$ square inches. This gives a stress on the bone of $\frac{2,520}{3.13} = 800$ pounds per square inch.

The forces involved in practice would certainly be much higher than this because the previous calculation was made for a femur with a medullary canal so small that it fitted the 0.25 inch width of the tip of the stem. In a medullary canal 0.5 inch in diameter the surface area available for bone contact would be at the most half the previous figure which would thus raise the stress in the bone to approximately 1,600 pounds per square inch (Example 6). This stress is of the same order as the compressive stress in the cortical bone of the cut end of the femoral neck calculated as 1,800 pounds per square inch in the previous example for a patient weighing 120 pounds.

Quite clearly the fretting movement between the stem of the prosthesis and the medullary canal would be enormous if load transmission were to be solely by "wedge-action" of the prosthetic stem, because the amplitude of movement in an axial direction would be seven times that of transverse dilation of the medullary canal. Very soon the prosthesis would sink into the medullary canal until the collar of the prosthesis made full contact with the cut end of the femoral neck and then "normal" conditions supersede.

Shear forces in load transmission—The preceding analysis of "wedging" of the stem of the prosthesis in a narrow medullary canal, where load transmission takes place from the medial and lateral surfaces of the stem of the prosthesis, is based on the assumption that the stem of the prosthesis is polished and that frictional forces acting tangentially to the surface are negligible. If, however, the metal surfaces are made rough, with asperities designed to key accurately into a large number of corresponding irregularities in the surface of the bone, very different mechanical conditions will exist when an axial load is transmitted from the prosthesis to the bone.

Here we must understand what is meant by the term "shear strength" of a material, and it is perhaps necessary also to comment that the word "shear" tends to evoke in the minds of orthopaedic surgeons the erroneous idea of evil or deleterious forces, since in orthopaedic surgery every effort is made to eliminate shearing forces and convert them to compressive forces when treating common fractures, of which that through the neck of the femur is the most outstanding. To the engineer a shearing force is merely a type of force; there is no connotation of it being either better or worse than compressive, tensile or twisting forces.

Let us consider the model illustrated in Figure 6 where an axial load is being transmitted to bone by a stem which has been provided with a multiplicity of serrations fitting intimately into the surrounding bone. These serrations can be considered as concentric rings on a shaft of circular cross-section such as would occur in a screw thread. In this construction there is no collar to transmit load to the cut end of the bone.

The profile of the serrations in this model is such that the load-bearing surface of each serration is normal to the axial load (the so-called buttress thread used for screws). This profile is used under the name of the "fir-tree" to anchor gas-turbine blades into the rotor where powerful centrifugal forces tend to pull out the blades.

It might be supposed that the pressure exerted on the bone by each serration could be reduced to almost infinitesimal amounts simply by increasing the number and the depth of the flanges, so as to increase the total area presented to the axial load. It is here that the concept of the "shear strength" of the bone must be appreciated, because the total strength of this

fixation is the force which would be necessary to shear off all the bony interdigitations lying between the flanges of the metal stem. In this example the plane of shear will be a cylindrical surface of length Y and of diameter X —that is, πXY square inches (Fig. 6). The total axial load which this bond will be able to resist before the bone fails in shear will be the product of the shear strength of bone (6,000–17,000 pounds per square inch) and the area of the surface exposed to shear.

One of the interesting features of this concept of shear strength in this connection is that, within limits, the number of serrations and the depth and shape of the serrations is unimportant as regards the ultimate strength of the bond, because the amount of bone exposed to shear is governed primarily by the length and external diameter of the serrated stem.

If we consider a prosthesis with a *parallel*, serrated, stem like that in Figure 6, 5 inches in length and 0.5 inch in external diameter, we have an area exposed to shear of 7.8 square inches. If we take the shear strength of bone as 11,000 pounds per square inch (the mean of 6,000 and 17,000 pounds per square inch) we see that such a bone could resist an axial load of $\frac{11,000}{7.8} = 1,410$ pounds (Example 7).

If we calculate the shear stress in the bone if a serrated stem were to have been used in Example 6, in which the data were taken from actual bone specimens, an interesting comparison is available.

A senile patient weighing 120 pounds exerts a resultant force of 360 pounds on the prosthesis (using a factor of 3 to allow for muscular and dynamic forces superadded). If a prosthesis is used with a serrated stem to present a surface area to shear of 7.8 square inches we see that the shear stress on the bone is only $\frac{360}{7.8} = 46$ pounds per square inch. If we take the shear strength of bone as 11,000 pounds per square inch we see that there is in this example a factor of safety of $\frac{11,000}{46}$ which is approximately 240 times (Example 8).

In other words, by using the surface of the stem of the prosthesis in shear it is possible to function with stress in bone very much below the shear strength of bone, whereas using a conventional prosthesis taking all the load on to the collar the bone is functioning much closer to its compressive strength, the factor of safety being reduced from 240 in the former to only 6 in the latter (Examples 4 and 5 above).

Load transmission by the use of cement—It is not a practical proposition to bond a femoral head prosthesis to the femur by means of a simple mechanical “fir-tree” design of prosthetic stem. This is because a fir-tree anchorage has to be inserted “sideways” into its bed, for it is obvious that if it were hammered down the medullary canal it would in fact be a “broach” and would cut away most of the bony projections. Attempts have been made by many surgeons to use prosthetic stems designed as screw threads but here anchorage against rotation is the difficulty.

It is here that the use of a quick-setting cement solves the problem. When “cold-curing” acrylic cement is used, the medullary canal is first broached to a size permitting the prosthesis to be tried in position as a loose fit. The cavity is then filled with the cement of dough-like consistency and the tapered (smooth) stem of the prosthesis is inserted to expand the cement and inject it into the trabecular spaces previously occupied by fat. When the acrylic cement has hardened it transmits the axial load from the prosthesis not by acting as an adhesive or glue but merely by making an accurate cast of the rough interior surface of the medullary canal. The preceding mechanical analysis has indicated that as regards the shear strength of the bone a large number of small irregularities is just as good as a smaller number of large irregularities. The cement acts as though it were an integral part of the stem and there is no need to provide mechanical irregularities on the surface of the metal stem; indeed to do this is most unwise as it will make extraction, should this ever be required, almost impossible.

The total surface area of the cement presented to the bone, and exposed to shearing forces under axial load, is obviously much greater than the surface of the metal stem. In this respect a very fortunate happening is that in elderly patients (and in those with rheumatoid arthritis) in whom the medullary canals are large and the cortical bone thin, the increased external diameter of the cement reduces the shear stress even below the 46 pounds per square inch estimated from the data given in Example 8. In an osteoporotic femur with a large medullary canal the cone of cement 5 inches long could be more than 1 inch in diameter proximally and 0.6 inch at the tip. This gives a surface area exposed to shear of 12.5 square inches. For an axial load of 360 pounds the shear stress on the interface between bone and cement will be $\frac{360}{12.5} = 28.8$ pounds per square inch. Taking the average shear strength of bone as 11,000 pounds per square inch this means that we are working within a factor of safety of $\frac{11,000}{28.8}$ which is approximately 380 times. It is therefore almost paradoxical that using cement in elderly patients (or patients with rheumatoid arthritis) where the medullary canal is much larger at the upper end of the femur than it is in healthy young adults, the factor of safety in the fixation, calculated in relation to the shear strength of the bone, is much greater than it would be in the narrow medullary canal with bone of the same shear strength (Example 9).

In this example, where the figures are taken from a senile patient, we see that the factor of safety using cement and the shear strength of bone is $\frac{380}{6} = 63$ times greater than the factor of safety using a conventional prosthesis and the compressive strength of bone.

Direct mechanical tests of cement fixation—That these calculations are supported by the facts is indicated in experiments reported in detail elsewhere (Charnley and Kettlewell 1965) in which femoral head prostheses were subjected to axial load in femora freshly removed from cadavera before and after they were cemented in position. Before cementing in position, using the standard technique of hammering the prosthesis into position with mallet blows equivalent to those used in the operating theatre on a living patient, a load applied axially caused a further migration distally of the prosthetic stem of four-hundredths of an inch per 100 pounds of load. After extraction and then fixation with cement the migration of the prosthesis under axial loads was on average no more than two ten-thousandths of an inch per 100-pound load. In one particularly interesting case, from a subject aged eighty-one years, the calcar femorale cracked under a load of 150 pounds when the prosthesis was inserted without cement; after it was cemented in position the same prosthesis tolerated a load of 380 pounds with the stem subsiding only about one-thousandth of an inch.

The fibrous bond—In this mechanical analysis the bond has been considered as taking place directly between the surface of the cement and the living bone. Histological examination shows, however, that after about six months a thin layer of fibrous tissue becomes interposed between the surface of the cement and the bony trabeculae. This layer of fibrous tissue varies in thickness from about 5 μ to about 250 μ , the thinnest layer being related to the distal half of the stem in the cortical tube of the shaft of the femur, and the thicker layers of fibrous tissue being related to the cement in soft cancellous bone in the trochanteric level and related to the proximal quarter of the cement. There seems no doubt that this indicates slight movement, probably from elasticity of the spongy bone of the trochanters, and suggests the idea that the most effective site of weight transfer is in the distal half of the cemented stem.

It is necessary to consider whether the interposition of a thin lamina of fibrous tissue materially alters the foregoing mechanical analysis. The existence of a layer of fibrous tissue, no matter how thin, always evokes the image of movement between an implant and bone, and for many orthopaedic surgeons the phrase "fibrous union" is synonymous with an inferior result.

The clinical fact must be remembered that these cemented prostheses do not become loose within the period of up to four years that they have so far been under observation (Charnley 1964). The theoretical analyses put forward are therefore not just speculations but are an attempt to provide an explanation of an established fact.

The essential point to be reconciled with the theory of fretting applied to the bonding of a prosthesis to bone is that any fibrous interposition, no matter how thin, must permit slight relative movement between cement and bone when the bond is loaded and unloaded. In cases which have been tested for fixation by re-exploration two and three years after the first operation no movement of the prosthesis was detected when manual force was used, though it is impossible to say whether micrometers would reveal a minute amount of resilience between the prosthesis and the bone.

It is probable that the clinical success depends on the roughness of the surface of the cement. The roughness of the cement results from the imprint of the freshly fractured spicules of cancellous bone in the soft cement at the time when the prosthesis was inserted. For some weeks after the operation the patient walks on a bond consisting of direct contact between cement and bony trabeculae, the tips of which have been killed by free monomer and by the heat generated in the "curing" process. Months later the dead cancellous bone is in part

FIG. 7

Model illustrating bond of cement A, to bone C, through medium of lamina of fibrous tissue B. Fibrous tissue is integral with C, but cement can be pulled away from fibrous tissue by a normal force 3. Under a tangential force 2 no relative movement occurs between A and B and all elastic deformation is within the substance of B.

absorbed and replaced by dense fibrous tissue and in part replaced by new trabeculae of bone. The facts indicate that the process of replacement is ahead of absorption, so that no loose phase intervenes in the transition. The rough surface of the cement retains a mechanical grip on the fibrous tissue and this type of mechanical hold is especially suited to resisting shearing forces between the cement and the surface of the fibrous tissue.

Whenever there is a tendency for two rough surfaces to slide one on the other the harder surface will cause the superficial layers of the softer surface to move bodily with it. This can happen even though no adhesion or "glueing" exists between the two surfaces, provided that the softer substance is intimately "keyed" into the surface asperities. A mechanical analogy is shown in Figure 7 where the fibrous intersection is represented by a layer of rubber B which has been cast into the surface roughness of the cement A which represents the prosthesis. The opposite surface of the rubber has been made to adhere by means of a glue to the surface C which represents the bone. If a force is applied to the surface to pull away the prosthesis A from the rubber B it will easily succeed in separating them, because they are only keyed together mechanically and there is no molecular adhesion between them. On the other hand a shearing or tangential force will move the cement of the prosthesis A a short

distance in relation to the bone surface C without any sliding movement at the cement/rubber interface between A and B. This shearing movement of the cement of the prosthesis A, relative to the bone C, takes place entirely as elastic deformation within the substance of the rubber. In other words, the existence of a thin elastic lamina in this situation will prevent the bone surface from being damaged by the cement in the process of "fretting," and movement is therefore possible without abrasion or attrition.

We see therefore that one of the reasons contributing to progressive loosening of prostheses such as the Judet or the Thompson is that their smooth polished surfaces cannot hold fibrous tissue; shearing forces will therefore cause relative movement between the metal and the fibrous covering of the bone. This is followed by abrasion and irritation of the fibrous layer with absorption of bone. It would not be sufficient to expect success merely by providing a roughened surface to the stems of conventional prostheses, because it is the cement in its plastic state which adapts to the bone surface and is responsible for creating the conditions in which the fibrous tissue takes over the fixation without a phase of fretting.

The physical quality of the fibrous tissue composing the anti-fretting lamina has not been mentioned. It might be suggested that fibrous tissue would be too "soft" or too "rubbery," to afford a firm bond. The thickness of the lamina will have a bearing on this concept because a thick layer would permit greater resilience than a thin layer, and it is fairly certain that the thinness of the layer encountered in these specimens is closely related to the success of the result. But it must not be forgotten that rubber only demonstrates its quality of elasticity when space is provided into which the rubber can expand from the compressed parts. Rubber in elastic deformity retains a constant volume and therefore in a confined space it is an elementary mechanical fact that rubber shows no elasticity against compression.

SUMMARY

It has been shown by mechanical analysis that by using cement to bond the stem of a femoral head prosthesis to bone two advantages are obtained when the conditions are compared with conventional methods.

1. "Fretting" between the implant and the living bone is eliminated. This source of persistent relative movement is probably the most important starting point for the progressive loosening of weight-bearing implants.
2. When cement is used the bond with the bone is exposed to stresses which are of an order three hundred times less than the shear strength of bone. The conventional prostheses expose the bond to compressive stresses which are near to the failure limits of the compressive strength of bone, especially in elderly patients with atrophic cortical bone in the femoral neck.

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