THE MECHANICAL PROBLEM OF THE ARTIFICIAL HIP

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"Just as the basic idea of Lane plates lived on by changing towards the more natural, so will arthroplasties involving interposition of inert materials survive only if they also tend toward the more natural." Bryan McFarland, 1954.

The complete solution to the problem of arthroplasty will be found only when we can provide our patients with new hips either made of, or able to unite with, living bone. Although it is by no means certain that such biomechanical synthesis is possible, this remains the goal for research, and unsuccessful attempts at welding artificial femoral heads and sockets in place by means of fast-setting dental acrylic have in fact been made by Haboush (1953). A more promising field lies in the covering of raw bony surfaces by grafts of articular cartilage which, nourished by synovial fluid, do in fact survive readily along with a thin plate of subjacent bone (Gallie 1956b). Unfortunately in the hip survival of the cartilage is nullified by changes due to vascularity or disease in the underlying femoral head, and it is perhaps because this abnormal bone is not removed that "cartilage-cup arthroplasty" has not been more successful. Some modification of it may provide the answer we are seeking.

Meanwhile, when in 1939 Smith-Petersen produced the first really practical artificial hip by means of a Vitallium cup, a strong fillip was given to the search. The articular surfaces, remodelled, with the cup between them, produced a brand new cartilaginous lining to the joint in response to the frictional and weight-bearing stresses on each side of the mould (Smith-Petersen 1939). As a result, the complete replacement of the femoral head by Judet followed as a matter of course.

When it became apparent that these acrylic replacements were unable to stay the pace of life in the human body for more than a few years there descended a fog of pessimism dense enough to obscure for a time the brilliance of many of the results during the earlier post-operative periods. As an example, Charnley (1956) considered mobility in the reconstructed hip an "unattainable ideal," and any form of replacement arthroplasty "doomed to failure" if only because of the physical differences between living tissue and inert substances. But whereas a few years ago the case for arthroplasty rested solely upon the fact that every other hip reconstruction was an admission of defeat, that case must now also rest upon the achievement of numerous successes with inert appliances.

Denham and Law (1957), in a recent analysis of two hundred cup and replacement arthroplasties, claimed four outstanding factors which, taken together, can be depended upon to produce—in eighteen months to two years—a successful and permanent cup arthroplasty. These are the use of a wide-mouthed cup of the Smith-Petersen type, an adequately excavated acetabular roof, intact abductor insertions and freedom from post-operative complications. But they have not found a dependable answer to the problem of those patients whose femoral heads are so unserviceable that they demand resection and replacement, or those whose age or general condition make them unsuitable for any method which will not allow early post-operative weight bearing on the new joint.

For an artificial hip, held in position by mechanical forces, there are two principal requirements: suitable materials and sound mechanics. The former will be discussed briefly later on; as to the latter, there is now strong evidence to support the contention that a prosthesis remains successful just so long as its mechanical fixation remains sound; so the next step in the evolution of hip arthroplasty ought logically to lie in the field which is most
within our compass—that of design. We must aim, as an engineering friend remarked in Irish vein, at putting back into the bone the stresses already there, by placing them along the lines already adapted for them. And we must also see that we do not create new ones.

THE UPPER END OF THE FEMUR

Architecture—The lines of stress which normally conduct the great forces of weight and movement through the femoral head and neck demand first consideration if it is accepted

![Diagram of forces](image)

**Fig. 1**
The mechanical stresses in the upper end of the femur have been compared to those of a lamp bracket and a crane.

that the primary function of a mechanically sound hip is to transmit those forces smoothly between acetabulum and femoral shaft. The more a prosthesis can be made to lie in balance with these natural lines of stress above and below it, the less intrinsic strength it will need—in other words the more it can conduct the less it will have to withstand.

![Images of bone structure](image)

**Fig. 2**
To show the trabecular structure of the femoral head and neck.

The internal architecture of the upper end of the femur consists of a series of Gothic arches, converging towards the head and with pillars resting upon the diaphysial cylinder. Besides these cancellous arches there is also that remarkable internal spur of cortical bone,
the calcar femorale, best developed at trochanteric level and representing an upward prolongation of the diaphysial cortex into the interior of the neck through the lesser trochanter (Debeyre and Doliveux 1955).

The cancellous trabeculae forming the arches are arranged in two main columns running up the superior and inferior aspects of the neck to intersect and decussate in a dense central wedge below the head, and thence fan out into a series of bony plates which meet the articular surface at right angles to its tangents throughout. This system was likened by Ward in 1838 to a lamp bracket, and by von Meyer (1867) to a crane (Figs. 1 to 3). It is apparent that any force applied to the femoral head is at once transmitted to the central wedge, and thence to the cylindrical diaphysis—the crane post—by means of the superior (tensional) trabeculae working reciprocally in conjunction with the inferior (compressive) ones. In a lateral view it will be seen that only a few of the lamellae on the anterior and posterior aspects appear to join the decussations in the central wedge, the majority distributing themselves so as to buttress the key area by arching their backs against it.

Between the two columns lies a structurally weak area of less dense bone, based upon a third column stretched between the trochanters. This is Ward's triangle, which in the analogy of the lamp bracket represents the cut-out centre of a plate. This area, through which passes the central axis of the neck and which must clearly have a bearing upon the etiology of fractures, will provide no secure anchorage for nail or prosthetic stem.

The trabecular pattern is explained in terms of Wolff's law, which states that the form of bone is produced in response to the stresses encountered; or, more simply, that the bony trabeculations follow the directions of the principal stresses (Haboush 1953). The classical example quoted to support this is the sloth's femur, devoid of any trace of trabecular differentiation since he lives upside-down, hanging from a tree. It is interesting that as early as 1931 Campbell, delivering the Robert Jones lecture at the New York Hospital for Joint Diseases on the subject of arthroplasty, showed radiographic evidence of restoration of bony structure and trabecular rearrangement "conforming to the lines of stress in the reconstructed joint." The architecture may thus be said to arise from a combination of all the principal stresses about a neutral axis which becomes their resultant, so that whereas the mechanical axis of the femur is said to be a line between the head and the intercondylar notch, it is around the anatomical axis in the centre of the bone that the forces are actually carried.

It follows from this that in coxa vara the anatomical axis shifts downwards, and under the influence of increased tensile stress the superior trabeculae hypertrophy. In coxa valga the shift is upward and the compression trabeculae bear the greater brunt. One may readily see this when watching a crane in action, and it has also been demonstrated by photo-elastic observance of the structural changes occurring in an acrylic model, which under stress becomes birefringent so that its molecules are rearranged according to lines of force visible under polarised light (Figs. 4 and 5).

All the evidence seems to confirm that Wolff's law and the theories of Ward and von Meyer hold broadly true for the upper end of the femur. The actual nature of the process whereby Wolff's law is put into effect has also received recent attention from Scott (1957) who conceived the mechanical basis of bone formation as intermittent trabecular pressure and tension with elastic recoil. Further small-scale detail is to be expected with the use of such methods as microradiography and the electron microscope.
THE MECHANICAL PROBLEM OF THE ARTIFICIAL HIP

In its crane-like structure lies the key to the meaning and purpose of the femoral neck. Just as a crane is designed to bring the load over the work, so the femoral neck is designed to place the erect human form upon the broadest possible base. In modern times there has been a tendency to decry the femoral neck as an area of inherent potential weakness—the

![Fig. 4](image)

Scale drawing of movable-jib crane to illustrate the forces in coxa vara and coxa valga.

French surgeons for instance have dubbed it the "porte-à-faux" of weight bearing—but I believe it is only when the architecture becomes abnormal that the neck becomes a source of weakness.

![Fig. 5](image)

Figure 5—Appearances of models undergoing constant pressure seen with polarised light. (After Pauwels, reproduced by Debeyre and Doliveux 1955.)

![Fig. 6](image)

Figure 6—Arteries of femoral head and neck. (After Judet et al. 1955.)

To examine its mechanics more closely: the neck consists of an external cortical component, analogous to a fixed-jib crane, supported by an internal cancellous strutting system made up of the decussating trabecular columns. According to Tobin (1955), weight bearing in the upper end of the femur is predominantly by cancellous bone, which is of equal
strength—weight for weight—to compact bone. However, as Charnley, Blockey and Purser (1957) recently pointed out, the cancellous system does not commonly persist throughout life, the femoral neck in the elderly being no more than a hollow tube filled with fatty marrow. When this occurs the normal crane system becomes reversed, so that the stresses passing through the lower half of the head (which is normally held up by the tension of the superior trabecular column) are converted to shearing ones, contributing to the liability of fracture.

It is plain, therefore, that the cancellous trabeculae cannot be relied upon to support an artificial hip. On the other hand it may be possible to replace these columns, and if so, their own structural pattern should provide the soundest guide to the design of the replacement.

**Blood supply**—This is outside the scope of the present paper, but since the problem is so essentially biomechanical in kind, and any prosthesis so dependent upon the living tissue around it, a few salient facts will be recounted here. The work of Trueta and Harrison (1953) and that of Judet et al. (1955) is well known and certain broad conclusions have been made.

The blood supply derives mainly from an anastomosis in the trochanteric fossa of the circumflex, obturator and inferior gluteal arteries. Thence vessels run proximally in the capsular attachments, branches entering the neck by perforating the cortex. The common pattern is shown in Figure 6, and it is apparent that the superior group of vessels predominates. These are the lateral epiphyseal arteries of Trueta and Harrison, which terminate by entering the bone at the cervico-capital junction. The arteries of the superior group supply nearly all the cancellous bone of the neck, and in replacement arthroplasty the line of section should therefore be planned to spare them whenever possible.

Although there is a free anastomosis between all the arteries around the neck, as well as between intra-osseous and extra-osseous vessels, the direction of the intracervical vessels, once they have entered the bone, is mostly downwards towards the diaphysis. The diaphyseal arteries, on the contrary, do not appear to make a similar contribution to the nourishment of the neck or head, as one might have expected they would.

Except at the epiphysial areas of the head and greater trochanter the vascular pattern is unrelated to the trabecular pattern.

Judet (1955) and his colleagues found that the blood supply of both head and neck was largely destroyed by cutting the femoral capsular attachments, whereas a less complete resection jeopardised the capital but left intact the cervical flow—a point that bears on technique in both cup arthroplasty and replacement arthroplasty.

Reaming the cervical stump to make a smooth bed for a prosthesis, and drilling the neck for the stem, have been found to cause extensive destruction respectively of extra-osseous and intra-osseous vessels. Laing and Ross (1952) demonstrated necrosis beneath a prosthetic stem two weeks after operation, and Judet found similar changes later under the weight-bearing segment of the rim. However, the fourteen-day findings of Laing and Ross should be treated with reserve because, as Collins (1954) remarked, considerable necrosis occurs normally after fracture or section of a bone, as a natural prelude to resorption and regeneration. In another femur examined two years after a Judet replacement arthroplasty it was found that good revascularisation had occurred, though with a pattern that bore no relation to the normal. Judet et al. (1955) considered that stabilisation of a prosthesis—clinically and radiologically evident—follows this revascularisation, and he stressed the potential danger of weight bearing before the process is complete. He pointed out that the area of cervical stump to receive the main brunt of the weight is the very area where revascularisation has to take place after operation.

Nevertheless good results—some promising to endure—have now been reported in replacement arthroplasty with very early post-operative weight bearing (Moore 1957, Stinchfield et al. 1957).
THE MECHANICS OF WEIGHT BEARING AT THE HIP

The four main stresses that fall upon the upper end of the femur are tensile, compressive, bending and shearing. Some rotational strain may also be thrown on the neck during flexion and extension, independently of weight bearing. Trabecular formation, according to Haboush, occurs only in response to stresses that are either tensile or compressive.

Figure 7 illustrates the well known static mechanics of standing on one leg. Alterations in body weight are thrice multiplied within the hip joint, and it seems probable that against such loads the difference between a prosthesis in steel or acrylic affects the hip no more than changing to a lighter or heavier overcoat.

![Diagram of hip joint](image)

**Fig. 7**

Weight bearing in one leg; \( b = \) body weight; \( g = \) gluteal pull.

If \( PC = 3FC \), then \( g = 3b \). Then total load at \( C = g + b = 4b \).

There are several ways by which the hip may be relieved of its burdens. First, by leaning towards the weight-bearing hip; this is the attitude of the hop-scotch player, and it reduces the discrepancy in lever length by bringing the centre of gravity nearer the head. Secondly, by using a walking-stick in the opposite hand, which Blount (1956) computed to relieve the joint of one-third of its load. Thirdly, by moving the greater trochanter farther outwards from the head, as occurs in coxa vara and in lengthening of the femoral neck. (It is important to see that although these conditions relieve the femoral head, they actually surcharge the neck. The reverse holds good for coxa valga and neck shortening.) Fourthly, by bringing the head nearer the midline—as by deepening the acetabulum—in order to equalise the levers. Denham (1956) calculated that a medial displacement of one inch ought theoretically to relieve the hip of half a hundredweight. Fifthly, by transplantation of the abductor insertions farther down the shaft, which will tighten them up and increase the length of their working lever, but in practice has proved a disappointing operation. Lastly, by moving the shaft inwards by displacement osteotomy until some of the weight is borne directly upon a new pelvic fulcrum farther medially. Unless displacement is sufficient to do this, as McMurray (1939) stressed, the osteotomy will have no mechanical effect, whatever its biological ones.

The femoral head, as already seen, is structurally adapted to receive weight at right angles to the whole of its articular surface; but such an arrangement will work only if there is perfect fit between the joint components. Debye and Doliveux (1955) estimated the pressure on a normal femoral head at 2 kilograms per square centimetre, and they stated that in an ill fitting joint this figure may be multiplied forty times, especially when a conical head makes with the acetabulum what they term "contact polaire." Furthermore, even in the normal hip the load is not evenly distributed. Lloyd-Roberts (1955), by intra-articular injection of wax, demonstrated the "pressure area," greatest in extension (the position of maximum stability) but always less than half the articular surface area, and diminishing as the joint moves away.
from the extended position. These observations give point to the second major effect claimed by McMurray (1939) for displacement osteotomy, namely alteration of the pressure area through the inward rotation of the upper fragment which occurs post-operatively owing to muscle pull—an effect which of course can in no wise take place if internal fixation is used.

Besides these static loads, it is equally necessary, and far more difficult, to assess the dynamic stresses accompanying movement, controlled as they are by endless combinations of co-ordinated—or unco-ordinated—muscle action. The presence or absence of limp, to take the most obvious, is certainly a major factor in the behaviour of the natural or artificial hip, whether due to gluteal insufficiency, to a loose prosthesis or simply to pain. What is lacking is a quantitative method of measuring the forces concerned. Anyone who doubts the measure of uncertainty which exists about muscle action should read Henry's (1957) account of gluteus maximus, which, quoting a description of the muscle by Leonardo da Vinci (1498), debunks nearly every function since attributed to it.

Thus to consider the hip joint merely in terms of its bony components, as a ball and socket, is a dangerous over-simplification. Nevertheless it is this very complexity of the functional anatomy that casts us back upon the bony architectural detail designed to serve its needs.

**PROSTHETIC MATERIALS**

Great progress has been made in finding materials which, though lacking the supreme slipperiness of articular cartilage (Charnley 1955), are tolerated by the body for long periods. Figure 8 shows the radiographic and photographic appearances of a mild-steel plate inserted by Sir Arbuthnot Lane forty-two years ago to immobilise a birth fracture, and recently removed by the author. The metal was absolutely firm in the bone, and Dr Scales, who kindly examined the materials microscopically, found no unfavourable bony reaction despite considerable surface corrosion of the plate and screws. This specimen illustrates the long-term

![Fig. 8](image)

*Fig. 8*  
Non-stainless steel plate inserted by Sir Arbuthnot Lane forty-two years before. It is absolutely firm, and there is no erosion of the surrounding bone.
answer to corrosion—that once the appliance becomes surface oxidised the process automatically ceases; the short-term answer is removal of the appliance. But with the development of non-ferrous alloys such as Vitallium and the clinical trial of the element Titanium (Leventhal 1951), substances reported to be totally free from foreign-body reaction, corrosion ceases to be a problem and we are left free to consider the mechanical effects of an inert prosthesis.

![Figure 9](image)

**Fig. 9**
Leakage of blood between acrylic and metal parts.

But first it is necessary to be sure that the inert material is of adequate strength for the job. Bell (1957) described bone as “nearly as strong as steel and much more elastic,” with a bending stress of 35,000 lb. per square inch—three and a half times that of hard wood. Martz (1956), who tested a number of internal prosthetic appliances and found them in every case far weaker than normal bone, also pointed out that the “fatigue-strength” of surgical steels is only equal to half their ultimate strength on the bench, and that the rest given to a prosthesis while the patient is asleep will not suffice to prevent fatigue fracture. We must therefore guard against inserting an appliance which—however well designed—is yet bound to succumb to fatigue if the patient lives long enough. Research on materials continues, and in the United States standards are being developed and applied to prosthetic appliances so that the surgeon may know precisely what strength of material he is using (Scales 1956).

Figure 9 illustrates one of the drawbacks of a composite prosthesis, in this case steel and acrylic. Staining can be seen where seepage of blood has occurred between the component parts.

**THE EFFECTS OF AN INERT APPLIANCE IN THE UPPER END OF THE FEMUR**

When a prosthesis is inserted into bone there occurs, depending upon the mechanical conditions, one of two quite contrary and opposing reactions. The first may be called remodelling, and represents an active, dynamic response in accordance with Wolff’s law. This favourable reaction continues as long as a prosthesis remains firm in its bed. The second, which is really pressure necrosis, occurs when the strain on the bone—either because of frictional movement or bad positioning relative to the forces encountered—reaches a figure, as yet unknown, where remodelling ceases to take place.

Collins (1953, 1954) investigated the reactions and histological changes around Smith-Petersen nails, where weight-bearing forces were not involved, and Judet prostheses, where they were. His findings, as well as the experiences of Moore (1957) and Lippmann (1957), establish that the natural reaction of trabecular hypertrophy in response to Wolff’s law is essential for any successful arthroplasty, and does in fact occur about a prosthesis lying in balance with the
natural forces; whereas pressure necrosis, "l'arthrose sous-prosthtique" as the French surgeons call it, represents an unnatural response to a foreign body placed athwart those forces so that the bone can no longer maintain equilibrium with the new conditions.

Radiologically it is not always easy to distinguish the early stages of these two reactions, because both may show increased density, and because all too frequently an early favourable response may change almost imperceptibly to an unfavourable one. However, the beginnings of clinical deterioration can be depended upon to coincide with the appearance of unfavourable radiographic changes.

At its first entry a nail or prosthetic stem shatters a few trabeculae, causing slight intra-osseous haemorrhage. During the repair of this damage the foreign body becomes encapsulated, first by fibrin and then by fibrous tissue in which bone is later deposited parallel to it. In twenty weeks a complete bony shell is present (Fig. 10); at two and a half years plates of cortical bone supported by a cancellous framework are found, and at five years fully developed Haversian systems. A Smith-Petersen nail examined five years after insertion lay tightly within the bone, its channel lined with a periosteal envelope half a millimetre thick, supported by fully vascularised bony plates parallel to its vanes and as dense as the normal cortex of the femoral neck; bone had also grown into the cannula for a distance of two millimetres (Collins 1953, 1954). Figure 11 shows a Judet arthroplasty three and a half years old which illustrates this normal response radiologically, and in which at the time the clinical result was virtually a normal hip.
The mechanical conditions that tend to perpetuate or to undo this natural bone reaction will now be considered in relation to artificial hips.

**CUP AND REPLACEMENT ARTHROPLASTY—GENERAL CONSIDERATIONS**

A cup satisfies the mechanical ideal of conducting rather than trying to absorb the stresses, because weight bearing remains on the femoral head as in the natural hip. And it liberates itself from the strains of motion by allowing free movement on either side of it. There is nothing to loosen because it is already loose. It is not surprising that breakage, though occasionally reported, is rare.

Smith-Petersen soon discovered that the bony surfaces in contact with a cup became covered with fibrocartilage, approximating in some areas to hyaline cartilage. Further careful work by Gibson and Williams (1951) confirmed this, the authors concluding that "tissue form and cellular type correspond to functional demands and that for the reproduction of articular cartilage compression plus gliding movement are a necessary combined stimulus."

This early work defines the essential object of cup arthroplasty, which is to promote the formation of a new joint by the body—not to provide a ready-made one as in replacement arthroplasty. And besides this difference of aim, the indications and limitations of the two methods also differ correspondingly, so that it is just to contrast rather than to compare them with each other.

Denham and Law's (1957) review, as well as the recently published results of Moore (1957), Lippman (1957) and Stinchfield et al. (1957), have enabled us to judge how far these original concepts of the cup and of the replacement prosthesis can be translated into clinical language. For each method, when well executed, two broad conclusions emerge from the findings: 1) that for cup arthroplasty a period of at least eighteen months is required before the new joint will function properly, but that once satisfactory it may be expected to remain so; and 2) that for replacement arthroplasty the result is a functioning joint which can be used for weight bearing in the immediate post-operative period, but that its liability to mechanical failure in the course of time is far greater. Thus the clinical findings vindicate the ideas behind the two methods: the cup, free from mechanical strain, basically dependent upon the biological response of the surrounding bones; the replacement arthroplasty upon sound mechanics and secure fixation.

From the mechanical standpoint conservation of the head is vastly preferable to removal. The great forces of weight and movement are concentrated and balanced in the head, which acts both as hub of flywheel and pinnacle of support, so that the more it can be preserved the more benefit will be retained from its specialised structure and supreme concentration of stabilised forces.

But the limitations of cup arthroplasty are determined by the condition of the head. If the latter has been badly enough damaged by trauma or disease it ceases to be a mechanical asset and becomes a biological liability. The advantage of replacement arthroplasty is that, having removed the pathological area, one has to deal with the constant pattern of normal anatomy instead of the inconstant pattern of disease. Its disadvantage is that the pulley and working stock of the femoral crane are sacrificed. Thus the mechanical and biological factors work in opposite directions in the two types of arthroplasty.

**THE SMITH-PETERSEN AND CRAWFORD ADAMS CUPS**

The Smith-Petersen cup, even when well fitted after reconstruction of both joint surfaces, is liable to move from one eccentric position to another (Fig. 12). Because of this the concentric cup was designed by Adams (Fig. 13) and is possibly the only artificial hip which is free from mechanical objection. Anyone who has used it must have been impressed by watching the movement of this cup after insertion—during flexion and extension the head rotating axially within the cup, while during all other movements cup and head move together as a single
member. This mechanical excellence produced, as expected, better short-term results, but over longer periods they disappointingly evened out with Smith-Petersen's. Adams (1955) attributed this to inadequate blood supply to the femoral head, and Denham (1956) reported cases of fracture of the femoral stump under the tight-fitting cervical part of the prosthesis. He concluded that a loose cup with a wide mouth is safest—a point that supports Judet's work on the ill effects of neck reaming. Thus the mechanical assets of the Adams cup have been obtained at the expense of biological ones; yet the fact that the speed of post-operative recovery has been found equal to that of the Judet prosthesis is further evidence that in the early stages of any arthroplasty the result depends predominantly on the mechanical conditions.

THE JUDET PROSTHESIS

It was the observation of the relief of pain after removal of the osteoarthritic head in Whitman's operation that led Judet and Judet (1950) to devise the prosthesis which put
replacement arthroplasty first upon the orthopaedic map. Their view of the head as the main source of coxalgia is reflected in their refusal to resect the capsule at operation. This is at variance with the opinion and practice of many British surgeons, but is supported by Stinchfield (1957), in whose series capsulectomy made "little or no difference" in cases of fresh fracture, avascular necrosis or osteoarthritis, though it was considered to have improved the results in patients with non-union. The argument remains unsettled. Although Lloyd-Roberts (1955) has produced evidence of pain of capsular origin, there is as yet no convincing explanation as to why an osteotomy below the hip will generally relieve coxalgia more completely than an arthroplasty with capsular resection, nor of Denham and Law's observation that trochanteric osteotomy appears to be of itself a cause of pain. Here, however, it is the mechanical properties of the prosthesis which are under consideration.

As already seen, it is not surprising that the single articulation provided by a newly fixed prosthesis functions at first much better than a cup which at the same stage can claim to be little else than a loose body in the hip joint. It is equally certain that the clinical deterioration of a once-successful Judet arthroplasty is the direct accompaniment and result of its failure to retain firm fixation in the bone. This painful fact has been observed too many times of recent years for it to be obscured by Denham's suggestion that an adequate abductor mechanism is more important than a completely firm prosthesis, though both must be considered essential factors for clinical success.

The weight-bearing forces that mainly affect this prosthesis (Fig. 14) are shearing ones, which act between head and stem and are resisted by the superior part of the rim, and bending ones, which act on the stem and are increased by driving the distal end across the weight-bearing axis into the lateral cortex, and greatly increased if the superior rim gives way. Tensile and compressive forces, magnified respectively in proportion to horizontal or vertical lie, also affect the stem, though less severely. Compression forces pass through the substance of the head, as in any prosthesis. Rotation strain will occur only if articular movement is not free, because the stem does not cross the axis of rotation by going down the shaft.

Figure 15 illustrates the familiar triple sequence of mechanical failure: rocking of the stem with widening of its canal, breakage of the superior rim and finally breakage of the stem; and it is exactly what one would expect from the interplay of forces. It will be seen that the portion of broken rim corresponds accurately to the characteristic flattened facet produced by weight bearing on the superior part of the head. The presence of the facet also illustrates that this prosthesis has been adequately anchored against torsional strain, for had any marked degree of axial rotation occurred the facet would either not have appeared at all or else would have occupied a much larger area (Fig. 16). In the former specimen the facet occupies rather more than one-third of the head—only a slightly higher proportion than the static "pressure area" on the natural head already referred to and described by Lloyd-Roberts (loc. cit.). That the mechanically weak axle-brake provided by the flutings of a Judet stem, insecurely anchored in Ward's triangle, can in fact prevent rotation suggests again that these strains are slight when the stem lies wholly within the neutral axis of the neck.

Judit and Judet always intended the stem to penetrate the outer cortex, so as to direct the appliance constantly towards a fixed point, and for further control of rotation. But there are serious mechanical objections. The axis of weight bearing is crossed and the bending
moment on the stem—now anchored at points on either side of it—is greatly increased, especially with its centre lying almost unsupported in the empty spaces of Ward’s triangle. Moreover, Haboush (1953) has shown that the stress-concentration at the edges of a cortical perforation will be many times greater than elsewhere. It is interesting that in Buxton and

Waugh’s (1954) cases shorter stems were used so as not to penetrate the cortex. Though no comparisons have been made it may be significant that in this series a high proportion of favourable bone reactions were observed radiologically.

The intrinsic strength of a Judet head in fresh bone was found by Debeyre and Doliveux (loc. cit.) to be just about half that of an intact femoral head, yet despite this the appliance has shown itself able to survive weight bearing so long as its fixation remains firm. Loosening,
however, produces additional strains which prove fatal to it, and as Denham and Law's analysis shows, its breakdown occurs most rapidly when the result of operation is good and the patient strong and active. The mechanical failure of the Judet hip may be summarised by saying that breakage occurs not so much because it is bearing the patient's weight as because its axis runs counter to weight-bearing forces which inexorably loosen it.

These drawbacks led Judet and Judet to introduce a modified prosthesis (Fig. 17). The rim has been removed, to avoid reaming, and weight bearing is transferred to a flat surface obtained by cutting the head obliquely at 60 degrees to the neck. When the neck-shaft angle is 125 degrees this head is inclined at 25 degrees to the horizontal so that its weight-bearing forces lie almost entirely in the direction of the compressive trabeculae. The eccentric stem still crosses the axis of weight bearing though the strains must be much reduced, and it still runs through the middle of Ward's triangle. The greater trochanter and the superior column of trabeculae are not utilised by this appliance. A brief account of results with this prosthesis has been published (Judet 1957). Clinical successes are reported as 72 per cent, against 67 per cent with the original type, and there was only a single case of broken prosthesis.

Following the breakages of prostheses of the Judet type there grew up, especially in America, a variety of more massive prostheses. In 1954 Shepherd reported thirty-seven in current American use, and in Europe there developed a similar tendency towards greater resection of neck and deeper penetration of shaft. On mechanical grounds, however, there is reason to doubt whether these bigger and heavier prostheses are likely to prove better ones.

**INTRAMEDULLARY PROSTHESES**

The first trend that became evident was to devise appliances with stems curved to follow the weight-bearing axis into the shaft. These are open to criticism on the ground that their stems, though soundly placed for weight bearing, lie athwart the rotational axis so that during flexion and extension torsional strain occurs at the junction of neck and prosthesis. Figure 18 illustrates torsional effect on a hip of this type, which has dislocated and rotated nearly 180 degrees within the femoral shaft. Figure 19 illustrates four of the various ways in which this vulnerability to torsional stress has been overcome, though always with an increase of bulk or complexity.

The Merle d'Aubigné prosthesis demands sacrifice of the whole femoral head and neck, and its stem penetrates almost to the knee. It has the advantage of resting upon the best vascularised and most resistant part of the bone. Its weight-bearing surface is flat and its area distribution maximal; thus in one sense its size is its principal quality. Merle d'Aubigné
(1954) emphasised restoration of normal neck length and when necessary correction of anteversion. It may be commented that good results can occur in arthroplasty without full restoration of neck length, just as there may be good function in the natural hip when the neck is shortened by injury or disease. Neck restoration, though clearly important and advisable whenever possible, is not one of the four vital factors found by Denham and Law to be essential for success. The retention of hip movement after "central dislocation" may be further evidence on this point.

In face of the clinical results, some of over five years duration, recently published by Moore (1957) and Lippmann (1957), it may seem presumptuous to criticise the method of either on theoretical grounds. In 153 operations Moore had not a single post-operative dislocation, despite early walking. Lippmann, whose prosthesis incorporates the unique feature of a free-running artificial joint for the head, reported ten of his twenty-seven patients without limp—an improvement in this respect on Moore's figures which showed some degree of limp in all but six out of fifty-five patients.
But once any prosthesis leaves the neck and enters the shaft it becomes a battle-ground for the powerful strains and leverages set up by the lower limb in motion, and hence its own strength must be greatly increased. Although shaft fixation must be used when the neck no longer exists, the middle part of the bone was described by Frazer as "a hollow, bony cylinder," and it appears improbable that a long metallic stem, however well shaped, will lie there long without play, even allowing for adaptations of the bone around it in response to Wolff's law. With knee and ankle joints locked, as when bending forward, the forces acting and reacting on the upper part of such a prosthesis are amplified like those at the end of a long screwdriver or the lever of a hand-operated capstan (Fig. 20).

**ANCHORED CUP PROSTHESSES**

 Appliances designed to combine the advantages of cup and replacement arthroplasty were described by Valls (1952) and Fitzgerald (1952), both consisting of a metal cup anchored by a central trifin nail.

 A cup requires less secure fixation than a complete replacement, for the forces which it transmits will automatically tend to hold it on the femoral head. In favour of fixation is the fact that the natural hip is single-jointed, which would seem to account for the early functional success of head replacements compared with cups. Against fixation is the loss of that stimulus towards the production of new articular cartilage provided by movement of the cup upon the underlying head. Although the trifin nail does prevent movement, such control would be more effective and stress-free if it were distributed peripherally instead of centrally, as for instance by such an arrangement as that shown in Figure 21 (b).
THE FEMORAL SLEEVE PROSTHESIS

Lenggenhager (1954), aware of the disadvantages of a central stem, designed an artificial head with a sleeve to lie over the neck like the crown of a tooth (Fig. 22). He pointed out that a cylinder will bear greater loads than a solid stem of equal mass and material, as well as achieve wider distribution of stress. Multiple perforations are provided to anchor the appliance by the growth of living tissue through them. Lenggenhager emphasised that the sleeve must fit the prepared cervical stump as tightly as possible and that it must be hammered on as far as the greater trochanter.

Although it is mechanically attractive, there are serious biological objections to this appliance. Lenggenhager objected to the malevolent effects of central stems on intra-osseous vessels, yet Judet et al. (1955) showed equally convincing evidence that important arteries entering the cortex are destroyed by reaming the neck—an essential pre-requisite for the sleeve prosthesis.

During a visit to Berne in 1956 I was able to discuss the results of Lenggenhager’s operation with a member of his team. The operation was initially a functional success, but in many cases it was complicated after a while by gross spur formation and osteophytosis round the base of the sleeve. A change-over from acrylic to metal made no difference to this, although it did prevent breakage. The Lenggenhager clinic has now abandoned the method in favour of an anchored-cup arthroplasty of the Fitzgerald type.

DISCUSSION

It is fair to say that cup arthroplasty has now established its place in orthopaedic surgery. With proper selection of cases and operative technique it can be depended upon to produce an enduring and satisfactory hip joint in somewhere about two years from the time of operation. This prolonged rehabilitation is a trivial price for a young or middle-aged patient to pay for relief of disablement, but one quite beyond the means of an aged person with—for instance—an ununited fracture. For this large group of patients the rapid return to painless weight bearing possible after replacement arthroplasty is what is needed. The remaining section of this paper is concerned with theoretical steps towards the design of a permanent prosthetic replacement, and it is submitted that success will be directly dependent upon mechanical soundness, which in turn will occur ideally with a prosthesis lying so intimately balanced that the surrounding bone will react naturally and dynamically to it according to Wolff’s law. In other words, not only must the bone support the prosthesis but the prosthesis must reinforce the bone. The following six propositions are submitted with these ideals in mind.

1. An inert artificial hip should aim at being a free conductor of natural forces with which it lies in equilibrium.
2. It should aim at being stress-free rather than stress-bearing.
3. The more it can be confined to the femoral head and neck, where weight bearing and rotation are concentrated naturally around a common axis, the more it will achieve these aims.
4. When there is a choice the level of section should be planned to spare the superior group of arteries at the cervico-capital junction.
5. The pressure of weight bearing should be distributed to as large an area of bone as possible along the lines of stress within the femur, and without weakening the cortex by perforation.
6. The design should aim at maintaining the length of the femoral neck, and at the same time allow for the possibility of absorption without interference with its mechanics.

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With regard to absorption, it is interesting that Charnley et al. (1957) found that in their cases of fracture treated by spring compression the loss of bone was always in the head. This suggests that absorption may be directly related to impairment of blood supply, and makes one feel that a prosthesis must not seek firmer fixation at the expense of blood vessels, for the more it can be kept from exerting pressure on these the better chance it will have of not producing absorption.

In the femoral neck—when the cancellous bone ceases to play its part owing to age—the crane system consists of three cortical components, welded together, but which may be considered separately in terms of a hip prosthesis designed to reinforce them. Thus the top of the shaft—the greater trochanter—is the top of the crane post, the superior cervical cortex the tensional component and the inferior cervical cortex the compression bar or jib. As already seen, the normal arrangements for conducting stress from head to shaft are to some extent reversed owing to the cancellous decussation below the head having ceased to function.

Figure 23 illustrates the angles of weight-bearing forces in the normal hip, taking Inman's (1947) average figures of 125 degrees for neck-shaft angle and 167 degrees for that made by the compression trabeculae with the shaft. This may be compared with the system of forces in the Lippmann prosthesis (Fig. 24), where it will be seen that a new crane, placed farther medially and using smaller leverages, has been substituted for the normal. The diaphysial cylinder is no longer the crane post, but a metal rod down its lumen; the greater trochanter is abandoned and the cortex twice perforated. On these criteria the Moore prosthesis (Fig. 19) comes nearer to imitating the natural system but has the drawbacks of great bulk, and of demanding sacrifice of almost the whole femoral neck.

Figure 25 illustrates a triangle of forces which might be expected to support a prosthesis placed at 30 degrees
to the horizontal—the angle of stability in fractures and also that at which the epiphysis is placed by nature during the middle years of childhood. This angle lies between that of the original Judet prosthesis (placed at 90 degrees to the neck) and that of its modification (placed at 60 degrees), and might enable some use to be made of tensional as well as compressive forces, working reciprocally within the bone as a replacement for the lost columns.

Though the foregoing seems theoretically the best mechanical arrangement, because of manufacturing difficulties and because of the need to allow more latitude for the descending limb of the prosthesis in relation to neck-angle variations, the experimental models shown in

![Diagram of hip prosthesis]

**Fig. 26**
Experimental models of a hip prosthesis based on the theoretical conceptions described.

Figure 26 have been evolved with the generous aid of Mr Maurice Down. (For full restoration of neck length a more globe-shaped head than this would be necessary, and for elective conservation of the superior arteries a much higher line of section.) The stage of clinical trial has not yet been reached. Recent history emphasises the words of Buxton and Waugh (1954) that "generalisations made from purely mechanical premises may be unreliable when applied to a particular patient," and the theories of this paper are set down as a target for further critical analysis. There are no conclusions.

**SUMMARY**

1. The structure and blood supply of the femoral head and neck, the mechanics of weight bearing, and the known effects of an inert foreign body are considered in relation to arthroplasty.
2. Some artificial hips are reviewed from the biomechanical standpoint.
3. From the information now available it is inferred that mechanical soundness and clinical success are not only co-related but interdependent; and that the mechanical problem of design offers most scope for further development at the present stage of our knowledge.
4. To this end six propositions are submitted.
5. A theoretical replacement arthroplasty, confined to the head and neck, in which breakdown of the component forces suggests that reciprocal use of both tensional and compressive loads might occur as in the natural femur, is described.

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