MECHANICS OF THE KNEE AND PROBLEMS IN RECONSTRUCTIVE SURGERY

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Static force transmission at the knee is analysed using measurements from radiographs showing the position of the line of body weight and also the bones of the knee in their correct orientation during function. With this technique it is possible to suggest values for a variety of forces acting at the knee.

During function the degree of knee flexion is not as important as the angle that the thigh makes with the vertical. The tension in the extensor mechanism is not the same above and below the patella. Failure to recognise these two features results in fundamental errors.

The patella, the effects of patellectomy and of forward displacement of the attachment of the patellar ligament are discussed. The importance of the transmission of force in the coronal plane is emphasised with particular reference to total knee replacement. It is suggested that small errors of geometry, as seen in the anteroposterior radiograph, can produce large changes of load.

The large number of prostheses suggests that there are still many unsolved problems in total knee replacement. The variety in design seems to reflect differing opinions on the mechanics of the joint. The constrained hinge of the living knee joint presents a much more complicated problem to reconstructive surgery than does the ball and socket joint in the hip; for abduction, adduction and rotational forces between the two longest bones in the body must be resisted by joint surfaces which are relatively narrow. The surgeon and the engineer must understand the significance of, and the difference between, the "line of body weight" and "leg alignment" (or "anatomical axis"—Maquet 1972); both are important in the design and technique of arthroplasty which must take account of the forces transmitted through the knee.

The shape of the joint surfaces and the constraints imposed determine the nature of the equilibrium between gravity, which acts upon the body, the reactive forces and the actions of muscles. The resulting forces are so balanced during function that force transmission in the femur is largely axial, while bending moments and shearing forces are small. Moreover the tendency to anteroposterior glide in the knee also is small. This arrangement ensures economy in bulk and weight of the long bones and allows simplicity in the shape of the bone ends.

With a given degree of knee flexion a patient may adopt many different postures, thereby completely altering the forces through the joint. To relate these forces to the degree of flexion is therefore of no value. As the angle which the femoral shaft makes with the vertical is directly related to the position of the centre of gravity of the whole body, this angle is much more important.

Since 1970 suitable radiographs have been examined to determine the angle between the femur and the tibia when viewed from the front. Only true anteroposterior radiographs of apparently normal adults were used. The average angle was 7 degrees, plus or minus 2 degrees. Differences in sex and height did not produce any consistent variation from this value. Recently, the long cassette employed for scoliosis has been used and has proved valuable. As the whole leg must be seen on one picture taken with a single exposure while the patient is standing, it is desirable to have a cassette with graded screens, fast for the hip and slow for the ankle. On this long film the tibiofemoral angle can be measured and the relationship of the tibial spines to the line connecting the centres of the head of the femur and the talus can be checked. Only radiographs of fully weight-bearing legs are of value.

The position of the centre of gravity of the whole body and its relationship to the knee is most important. Braune and Fischer (1889) located the centres of gravity of the individual parts of the dismembered body; from their figures, the centre of gravity of the whole body can be calculated for different postures. But with a living subject, provided he is standing still, the line of body weight can be found, since in equilibrium it passes...
vertically through the feet. The smaller the area of contact with the ground the greater is the accuracy with which the line of body weight can be located.

FORCE TRANSMISSION AT THE NORMAL KNEE

A method similar to that used in the hip by Osborne and Fahimi (1950) has been used in Portsmouth since 1961 to assess the function of the knee before and after surgery (Denham 1959, 1963). Since 1974 the same method has been used to estimate many indices in the knee including: the shearing force, the axial compressive force and the bending moment in the femur; tension in the patellar ligament and in the quadriceps tendon; and patellofemoral and tibiofemoral reactive forces (Bishop 1977). A three-dimensional force analysis is needed to give a complete picture of static force transmission in the knee. In this paper we restrict attention to a two-dimensional analysis noting, however, that the true forces will generally be somewhat larger than those quoted. Experiments were performed on subjects whose knees were apparently normal and for whom it was reasonable to expect that equal weight was being taken through each leg.

If a subject is standing, taking weight equally on both legs, balanced upon a narrow rod of trapezoidal cross-section, the centre of gravity of the body is directly above the upper surface of the support (Fig. 1). If a radio-opaque plumbline intersects the upper surface of the support and casts its shadow on an x-ray plate held between the knees, distances between this shadow and various points on the lateral radiograph can be measured and moments taken. Figure 2 shows a radiograph obtained in this way.

It is possible to take a tracing of the radiograph and to draw on it the line of the extensor mechanism. In the example shown in Figure 3 the direction changes twice, once at the patella and once at the anterior part of the femoral condyle. When two surfaces are in contact and there is no friction between them the force of reaction is at right-angles to the surfaces at the point of contact. Thus it is possible to draw on this tracing the direction of $P$, the patellofemoral reaction, and $R$, the force resulting from the change of direction of the extensor mechanism at the femoral condyle (Fig. 4). It is not reasonable to assume that the tibiofemoral reaction is normal to surfaces of the bones, if only because of the presence of the cruciate ligaments. The reaction can be thought of in terms of an equivalent normal component $T$ and a tangential component $C$ acting at point $A$.

Moments may be taken about $A$ of all the forces acting on the lower leg. If $w$ is the weight of the lower leg acting through its centre of gravity at a distance $d$ to the right of point $A$, and $H$ is the supporting force at the forefoot (that is, the force of half the body weight), and if by direct measurement from the tracing of this particular radiograph, the shortest distance between point $A$ and the patellar ligament is 4.5 centimetres, and the horizontal distance between point $A$ and the line of body weight is 16.7 centimetres then, as will be seen from Figure 4,

$$4.5L + wd = 16.7H.$$  

Here, $L$ represents the tensile force in the patellar ligament. If, on the basis of information obtained from Braune and Fischer, the assumption is made that the weight of the lower leg is $H/6$ acting at 16.7 centimetres to the right of point $A$ then,

$$L = \frac{16.7H}{4.5}(1-\frac{1}{6}) = 3.1H.$$  

Alternatively, had the assumption been made that the value of $d$ was 10 centimetres instead of 16.7 centimetres, the calculated value of $L$ would only alter by about 8 per cent. Thus the value of $L$ varies but little despite wide differences in $wd$. 

THE JOURNAL OF BONE AND JOINT SURGERY
Simultaneous electromyograph tracings (Fig. 5) show that in balanced equilibrium the extensor effect upon the knee is not greatly affected by actions in the hamstrings or the gastrocnemius. Only the occasional burst of activity which helps to maintain balance is seen in these muscle groups, so that their effect can safely be disregarded in calculations of static force transmission.

The choice of point A for the purposes of calculation has the advantage that the collateral and cruciate ligaments cross the knee close to it, so the error in the value of L incurred by omitting possible tensions in these ligaments is small. The spiral of Fick (1910) illustrates how the centre of curvature of the articular surface of the femur changes position as one examines the bearing surface from front to back. If these changing positions are used in calculations of static force transmission, measurements are complicated, whereas the use of contact point A facilitates measurements. It is interesting to note that when the knee is examined as a whole, movement of the tibia over the femur is as shown in Figure 6.

It is found by measurement that the forces L and T in Figure 4 both act at 44 degrees to the horizontal. The forces which produce equilibrium of the lower leg may be resolved. In the vertical direction,

\[ T \sin 44 + C \cos 44 + w = L \sin 44 + H. \]

In the horizontal direction,

\[ T \cos 44 - C \sin 44 = L \cos 44. \]

It is found from these two simultaneous equations that \( C = 0.6H \) and \( T = 3.7H \).

The patella can be isolated as a system. For equilibrium to be possible, the forces \( P, U, L \) must act through a single point, \( U \) being the tension in the quadriceps tendon. Knowing the direction and value of

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**Fig. 4**
A tracing of one "line of body weight" radiograph showing the positions of the forces acting at the knee.

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**Fig. 5**
In a subject who was crouching on a narrow wedge, simultaneous electromyograph recordings were made using a twin-track recorder, first of the action of the quadriceps and hamstrings, and then of the gastrocnemius and soleus. Great activity is seen only in the quadriceps and soleus.

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**Fig. 6**
When the tibia and femur are connected by their ligaments, a pencil in the tibia traces an arc of a circle as that bone flexes and extends on the immobilised femur.
L and knowing the directions of the other two forces, we can draw a triangle of forces. From this the value of P is found to be 5.9H and that of U to be 6.0H (Fig. 7).

It is widely held that, since friction between the patella and the femoral condyle is very small, the tension in the patellar ligament is equal to that in the quadriceps tendon. This belief is not compatible with calculations, nor with experimental findings. Results in cadaver experiments (Bishop and Denham 1977) in which the position of the patella and the direction of the tendons above and below were reproduced as accurately as possible, confirmed this. During normal function tension in the patellar ligament is often substantially less than tension in the quadriceps tendon immediately above the patella. Calculations which ignore this fact are inaccurate.

As the extensor expansion passes over the anterior part of the femoral condyle it changes direction and a reactive force R acts on the bone (Fig. 8). As no sesamoid bone is present at this point, the direction of R bisects the angle made by the upper and lower parts of the tendon and, since friction can probably be ignored, a triangle of forces can be drawn. This shows that when U and U' are 6.0H, R is 1.9H.

Treating the lower end of the femur as a new isolated system and knowing the important forces which act upon it in the vicinity of the condyles, we can next calculate the value and direction of the force in the femur (Fig. 9). The exact position and direction of this force is not certain, so an axial force F has been drawn at a distance of y centimetres from the centre of the bone, and a shearing force S added at right angles to the femoral shaft. Using the values for C, T, P and R which have already been calculated and knowing the direction of F and S, a polygon of forces can be drawn to represent their action in this plane. By measurement of this polygon it is found that F = 6.7H and S = 0.4H.

Taking moments about point B of all the forces acting on the femoral condyle and remembering that the resultant moment for equilibrium must be zero if equilibrium prevails, we find that the axial force F of 6.7H acts 1.2 centimetres from point B. This is not quite in the centre of the bone so the axial compression force is accompanied by a very small bending moment in the femur.

These calculations were repeated using tracings of radiographs taken of three patients in postures with different angles of the femur. In this way a series of graphs were obtained (Fig. 10). These suggest that between the vertical position and 20 degrees of inclination of the femur, the patellofemoral reaction, the tension in the quadriceps tendon and in the patellar ligament, and the axial compression force in the femur rise slowly. From 20 degrees to 60 degrees there is a rapid rise to between five and ten times body weight. After 60 degrees these forces do not increase much. During the whole range of movement, the shearing force at the lower femoral epiphysis and the tension in the lateral and cruciate ligaments of the knee are never much greater than half body weight in each knee, and the maximum tendency to disrupt the joint is sustained at a time when between six and eight times this force is pressing the slightly curved surfaces together.

THE PATELLA

The patella plays an important part in distributing to the femur large forces that originate in the quadriceps. The patellofemoral pressure is distributed through its thick resilient cartilaginous pad over a wide area which changes in contour in different degrees of flexion. It is not so widely appreciated that the patella exerts a great influence on the directions in which forces are transmitted at the knee. The direction of the patellofemoral thrust influences the reaction T and shearing C at the tibiofemoral contact A, the compression F and shearing

Fig. 7
If the directions of the forces U, P and L acting upon the patella are known, and if L is 3.1H, then the values of U and P can be measured on the force triangle.

Fig. 8
From the force triangle of U, U' and R, the value of R can be measured. The system in equilibrium to which the triangle relates is the short length of tendon that is in contact with the anterior part of the femoral condyle.

Fig. 9
The polygon of forces acting on the femoral condyle can be used to find the value of F and S, the axial thrust and shearing force respectively.
S in the thigh, and also the quadriceps tension $U$. Any change in the direction of patellofemoral reaction caused by a high, a low, a worn or an absent patella is likely to provoke abnormal stresses in the joint.

Calculations strongly suggest that when the knee is bent, patellofemoral reaction can be substantially greater than tibiofemoral reaction. In joint replacement, where the articular cartilage of the patella is badly worn and unable to adapt to a changing contour, we believe that a wide, smooth regular femoral surface is essential. This design factor is neglected in many knee prostheses.

The patella is a mechanical aid to the function of the knee, for the thickness of its bone and articular cartilage displaces the extensor tendon forwards away from the point of contact between the tibia and the femur (point A). Figure 11 is a tracing of Figure 4, but without the patella. Provided the configuration of the remainder of the knee remained unchanged, tension in the patellar ligament would be increased by about 30 per cent. After patellectomy the direction of the thrust of the extensor mechanism as it passes over the femoral condyle would bisect the angle between the line of quadriceps action and the line of the patellar ligament. This change influences the value and direction of other forces acting at the knee joint. With this subject in the position shown, patellectomy could result in about a 14 per cent increase in tibiofemoral reaction and a 250 per cent increase in the tangential force in the tibiofemoral joint.

Figure 12 shows the results of tests performed on a patient who, following a severely comminuted fracture, had had a patellectomy four years previously. Even though the clinical result of the operation seemed to be successful, the graphs show that function in the damaged leg was not as satisfactory as it seemed and severe strains were being placed on the opposite knee. The reaction at the sole of the foot shows that the knee without a patella was not supporting its fair share of the load.
FORWARD DISPLACEMENT OF THE ATTACHMENT OF THE EXTENSOR MECHANISM

Maquet's operation of tibial tubercle advancement seems a rational way of reducing some of the forces acting at the knee (Maquet 1972). Figure 13 shows that, provided other factors remained the same as in Figure 4, there could be a 19 per cent reduction in the tension in the patellar ligament if the tubercle were displaced forwards by one centimetre. With the opening of the angle which the extensor mechanism makes as it passes over the patella, patellofemoral reaction and tension in the quadriceps would be reduced. The orientation of the patellofemoral thrust would be changed however. This would slightly increase tibiofemoral reaction and the shearing force in the knee joint.

Forward displacement of the patella or forward displacement of the tibia upon a tibial prosthesis or osteotomy, could result in a decrease in tension in the patellar ligament with the accompanying changes described above. Osgood-Schlatter disease, with its forward displacement of the tibial tubercle, would lead to a reduction in forces transmitted at the knee.

LIMITATION OF KNEE EXTENSION

Full extension of the knee usually includes a few degrees of hyperextension. This "over-centre" mechanism is an advantage, for in "full extension" the extensor muscles can relax without the leg giving way. In this position the centre of gravity of the body is situated in front of point A and all ligaments except those of the extensor mechanism are tight. The tibia and femur have rotated upon each other and reached a configuration in which there is a large area of articular contact. Sudden collapse after a tap on the back of the knee of a subject who is standing in a relaxed posture is due to flexion of the joint past the tibial top dead centre.

In knee replacement some attempt should be made to restore lateral ligaments to their correct length, but it is of the greatest importance to ensure that the posterior ligament is preserved with its three sections intact and at the correct tension when the limb is in the normal fully extended position. This is done by removing just the right amount of bone and replacing it with an implant which, with its cement, is of the correct height. Preservation of this limiting mechanism is essential with an unconstrained total knee prosthesis. Moreover, since the posterior ligament is sensitive (and therefore capable of giving warnings of excessive loading) its retention must be better than the provision and use of metal or plastic stops in a constrained hinge total knee prosthesis.

LATERAL STABILITY

In the knee, static force transmission in the coronal plane—at right angles to the normal plane of movement—depends upon the width of the joint, the shape of the articular surfaces and the force holding these surfaces together. This latter force is determined by the body weight, the tensions in flexor and extensor muscles and the tensions in the relevant ligaments. The flexors and extensors normally control the degree of flexion by acting differentially so that movement occurs when unequal effects are produced by the two muscle groups. It is suggested that, pathological conditions apart, normal ligaments can play only a limited part in maintaining the integrity of the joint; ligaments stretch excessively only when muscle action does not protect them. Stability is easily achieved with normal anatomy and normal function; but in degenerative joint disease,
in the presence of weak muscles, or with violent lateral strains, the intrinsic stability of the joint with its congruous surfaces held together by the compressing action of both flexor and extensor muscles may be insufficient to maintain alignment. Too great a call is then made on the ligaments in resisting the disrupting action and they become stretched.

This may be illustrated by the following argument. In the absence of actions by the flexor or extensor muscles, the medial condyle in Figure 14 would carry the weight of the subject, the lateral condyle would lose contact with the articular surface and a net moment would have to be supplied by ligaments. If the hamstrings or quadriceps act as if to close the joint, however, there remains a differential loading of the condyles, but both transmit compressive force and there is no extraneous moment supplied by ligaments. Quantitative accuracy is difficult since moments must be calculated whose “arms” are comparable with the dimensions of the possible contact areas.

The statics may be studied with the help of anteroposterior radiographs showing the line of body weight. Figure 15 shows the great change in posture of the whole body which is needed to vary the position of the line of weight. Even though the subject has leaned over as far as possible, first in one direction and then in the other, she has only changed the position at which the line of body weight intersects the knee by 8 centimetres. This small variation, which is under voluntary control, contrasts with the very large distances which result from deformity in the knee. Even in the presence of minor varus or valgus deformity it is difficult to maintain the centre of gravity over the surfaces of the joint. With severe deformity this may not be possible and deterioration increases rapidly.

Viewed in the anteroposterior direction, the knee reveals an important kinematic (geometric) feature, namely that very small errors in the angle between the articular surface and the long axis of a bone cause great changes in the position of the line of body weight, relative to the knee. In the normal leg, the tibial plateau is at right-angles to the shaft of the tibia, while the femoral shaft is usually in 7 degrees of valgus. A variation of as little as 4 degrees in either of these angles can move the line of leg alignment from its optimum central position to the lateral third of the joint. A variation of 10 degrees can place it beyond the articular surface. In this deformed position the whole of the force transmitted by the tibiofemoral joint passes through one tibiofemoral compartment, and muscle action may be unable to close the contact in the other. The force transmitted in the one compartment is the sum of the body weight, the combined flexor and extensor muscle action and also the excessive tension which the ligaments suffer in maintaining the articulation. The worse the deformity the greater must the tensions be if subluxation is to be prevented. These protecting forces further increase pressure in the joint. With this vicious circle the rate of deterioration in the joint increases rapidly. In an attempt to move the centre of gravity of the body back to its normal position the patient will lurch to one side.
during walking, but the degree of voluntary control is small and other joints are strained.

It thus appears to be a first essential from the mechanical point of view that fixation of new joint surfaces to the long bones shall be at the correct angle if long-term success is to be achieved. In severe degenerative joint disease the leg is often fat, and accurate alignment is difficult. Intramedullary splines fixed to the new joint surfaces or removable pins which pass through the component at the correct angle simplify this part of the operation. With normal alignment of the leg, both joint surfaces are at right-angles to the line of gravity, so the tendency to lateral subluxation is small and the design of the implant can be simplified (Fig. 16).

CONCLUSIONS

This paper has only dealt with static force transmission and it is natural to enquire to what extent this is a serious limitation. It seems to the authors that the restriction may not be as great as it might appear. By static analysis it is possible to gain some idea of the maximum forces that can be exerted at the knee. For example, when the femur is in its most nearly horizontal position, the patellofemoral reaction is about ten times the half body weight. We suggest it is unlikely ever to be much greater than this. With the intervention of inertia forces the subject will ensure, by adjusting his posture, that this factor will not be exceeded. We may thus have a general picture of what the greatest forces are likely to be under dynamic conditions.

Voluntary control of the position of the centre of gravity of the body by changes in posture is a most important factor in function. Past experience is combined with the present athletic performance of which the subject knows he is capable, and the distances, loads and the time available are computed by the brain to give the correct posture for a particular action. It can be shown that leaning forwards a few centimetres can halve the force passing through the knee. Variations in the position of the line of body weight provide a simple and easily controlled way in which the subject can reduce the stresses in a joint during activity.

Using the techniques described in this paper and making certain assumptions in the manner of engineering analysis, it is possible to examine the effects of osteotomy, joint replacement and certain other operations. The results of this work will be reported in due course.

REFERENCES