Biomechanics of the Hip

This section is not intended to be a comprehensive analysis of the forces acting on the proximal femur and the acetabulum. The reader is referred to the exhaustively complete analysis of this subject by Frederick Pauwels in Biomechanics of the Locomotor Apparatus (7). It is, however, important to the success of total hip arthroplasty that one understands the factors influencing both the direction and magnitude of forces acting upon the femoral head. The forces exerted on the hip have their biological expression in the form of the femur and acetabulum, particularly in the location and orientation of the trabecular pattern. The forces exerted on the prosthetic femoral head in a properly performed total hip replacement will be very similar in both direction and magnitude.

Of all the species in the animal kingdom, only birds and man habitually use a bipedal gait. Even the larger primates use a quadripedal ambulation mode for most of their activity. When the weight of the body is being borne on both legs, the center of gravity is centered between the two hips and its force is exerted equally on both hips (Figure 1.23). Under these loading conditions, the weight of the body minus the weight of both legs is supported equally on the femoral heads, and the resultant vectors are vertical.

When the hips are viewed in the sagittal plane and if the center of gravity is directly over the centers of the femoral heads, no muscular forces are required to maintain the equilibrium position, although minimal muscle forces will be necessary to maintain balance. If the upper body is leaned slightly posteriorly so that the center of gravity comes to lie posterior to the centers of the femoral heads, the anterior hip capsule will become tight, so that stability will be produced by the Y ligament of Bigelow. Therefore, in symmetrical standing on both lower extremities, the compressive forces acting on each femoral head represent approximately one-third of body weight (7).

In a single leg stance, the effective center of gravity moves distally and away from the supporting leg since the nonsupporting leg is now calculated as part of the body mass acting upon the weight-bearing hip. Since the pillar of support is eccentric to the line of action of the center of gravity, body weight will exert a turning motion around the center of the femoral head. This turning motion must be offset by the combined abductor forces inserted into the lateral femur. In the erect position, this muscle group includes the upper fibers of the gluteus maximus, the tensor fascia lata, the gluteus medius and minimus, and the pyriformis and obturator internus. The combined resultant vector of the abductor group can be represented by the line of action M in Figure 1.24. Since the effective lever arm of this resultant force (BO) is considerably shorter than the effective lever arm of body weight acting through the center of gravity (OC), the combined force of the abductors must be a multiple of body weight. The vectors of force K and force M produces a resultant compressive load on the femoral head that is oriented approximately 16° obliquely, laterally, and distally. The orientation of this resultant vector is exactly parallel to the orientation of the trabecular pattern in the femoral head and neck (Figure 1.25).

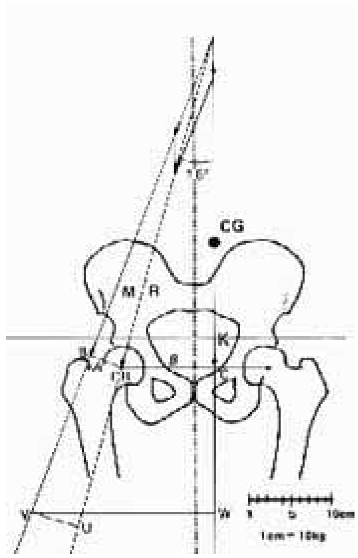

Figure 1.24. Forces of the hip in single leg stance. G, Center of gravity; M, muscle forces; K, effect of partial body weight; R, resultant vector. [Redrawn from Pauwels F. Biomechanics of the Locomotor Apparatus, Springer Verlag, New York, 1980 (7).]

Figure 1.25. A-P x-ray of a normal hip showing the compression trabeculae oriented parallel to the resultant compressive load on the femoral head.

The effect of this combined loading of body weight and the abductor muscle response required for equilibrium results in the loading of the femoral head to approximately 4 times body weight during the single leg stance phase of gait. This means that in normal walking the hip is subjected to wide swings of compressive loading from one-third of body weight in the double support phase of gait to 4 times body weight during the single leg support phase. The factors influencing both the magnitude and the direction of the compressive forces acting on the femoral head are 1) the position of the center of gravity; 2) the abductor lever arm, which is a function of the neck-shaft angle; and 3) the magnitude of body weight. Shortening of the abductor lever arm through coxa valga or excessive femoral anteversion will result in increased abductor demand and therefore increased joint loading. If the lever arm is so shortened that the muscles are overpowered, then either a gluteus minus lurch (the center of gravity is brought laterally over the supporting hip) or a pelvic tilt (Treridelenburg gait) will occur.

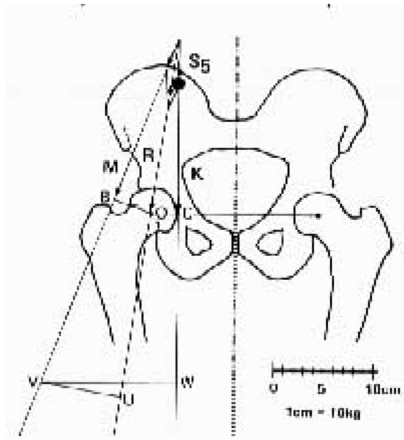

Figure 1.26. Forces on the hip with sideways limping. Note the reduction of vector M and R even though K is unchanged. R is also more vertically oriented. [Redrawn from Pauwels F. Biomechanics of the Locomotor Apparatus, Springer Verlag, New York, 1980 (7).]

Since the loading of the hip in the single leg stance phase of gait is a multiple of body weight, increases in body weight will have a particularly deleterious effect on the total compressive forces applied to the joint. The effective loading of the joint can be significantly reduced by bringing the center of gravity closer to the center of the femoral head (Figure 1.26). Sideways limping, however, requires acceleration of the body mass laterally, its deceleration during the stance phase of gait, and then its acceleration back to the midline or even to the other side as the single leg stance phase changes to the opposite extremity. This requires considerable energy consumption and is a much less efficient means of ambulation than the normal situation in which the hip is subjected to these considerable forces. Another effect of sideways limping is that the resultant vector becomes more vertical because the center of gravity is acting in a more vertical direction, and therefore the bending moment the femoral neck is increased.

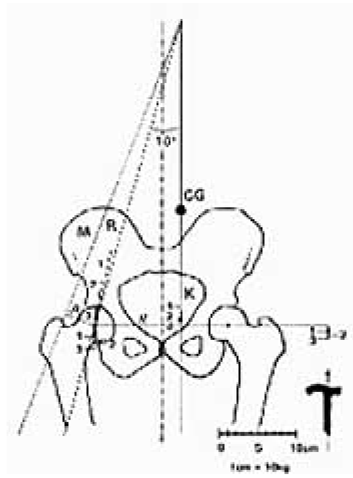

Figure 1.27. Forces on the hip with the use of a cane. Force levels and vectors correspond to the magnitude of pressure referred to in Table 1.1. [Redrawn from Pauwels F. Biomechanics of the Locomotor Apparatus. Springer Verlag, New York, 1980 (7).]

Another mechanism for reducing the resultant load on the femoral head is the use of a walking stick in the opposite hand. Since some of its force is transferred to the walking stick through the hand, the effective load of body weight is thus reduced in two ways: 1) the effective load of body weight is reduced; 2) since the turning moment around the femoral head is reduced, the abductor demand is also reduced (Figure 1.27).

TABLE 1.1. Influence of a Walking Stick on Forces across the Hip

	Pressure of stock (kg)	Static load across the hip(kg)	Angle in inclination from the vertical of the compression force on the femoral head
R	0	17.5	16°
1	9	100	13°
2	15	51.2	8°
3	17.5	30.26	0°

Adapted from Pauwels F. Biomechanics of the Locomotor Apparatus. Springer Verlag, New York, pp 1-228, 1980 (7).

Pauwels (7) has calculated both the total compressive load on the femoral head and the angle of inclination of the vertical compressive loads for different forces applied to the walking stick (Table 1.1). It can be that only 9 kg of force applied to a cane in the opposite hand reduces the load on the femoral head by nearly 40%. The same effect could also be achieved by a 40% reduction in body weight. Also the angle of inclination with this degree of unloading is not significantly different from normal, so that using a stick to unload the femoral head produces lower bending forces around the femoral neck than sideways limping. Therefore, in the rehabilitation of patients after total hip arthroplasty, the use of a stick to prevent sideways limping is always preferable.

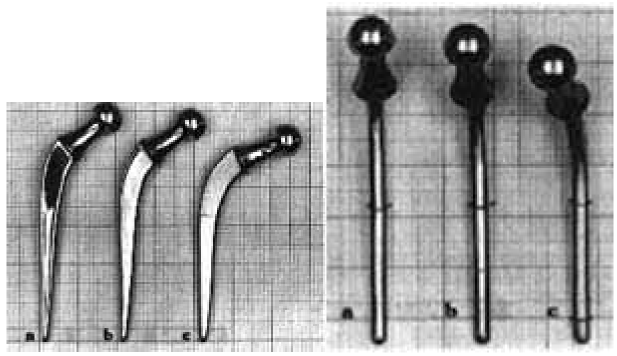

Figure 1.28. A-P and lateral photographs of three Charnley femoral stems: unused stem (a), failed prosthesis (b), fatiguing prosthesis (c); b and c are deformed in varus and retroversion.

The form of the femur and the orientation of the trabecular pattern in the proximal femoral metaphysis and epiphysis would support the conclusion that the principal loading of the femoral head is in the coronal plane. However, there is another manner of loading that also has clinical relevance to total hip arthroplasty and may also play a significant role in loosening. When an individual rises from the seated position or climbs stairs, the forces of body weight are applied to the anterior surface of the femoral head. The femur itself is prevented from rotating in response to this applied load by the stabilization of the posterior femoral condyles against the tibial plateaus. In addition the psoas tendon inserting into the lesser trochanter prevents this applied load from rotating the femur internally. This anteriorly applied force therefore produces a twisting strain on the proximal femur. That this must be so is demonstrated in two Charnley total hip femoral stems that were recovered after failure through loosening. In both instances the distal portion of the prosthesis remained fixed in the diaphysis while the proximal cement mantle loosened. Although both specimens had deformed into varus, they both also had deformed more in retroversion (Figure 1.28). The more deformed of the two specimens was from a 40-yr-old postal worker who had a total hip replacement for avascular necrosis and returned to work as a postman, which required frequent squatting and lifting of packages.

This aspect of loading of the proximal femur takes on particular importance for femoral stem design since anteriorly applied loads will produce a twisting strain on the stem within the medullary canal. Vertical loading of the femoral component will produce compressive load on the medial side of the femoral stem and tension loads on the lateral side of the stem, whereas anterior loading will produce shear stresses at the prosthesis-bone-cement interfaces. Since smooth stems are capable of transmitting load only in compression, this latter mode of loading is an argument for fixation that has the capability of transmitting all three mechanisms of stress: compressive, tensile, and shear. It also implies that it is inadequate to analyze the validity of femoral stem design by only simulating vertical load and that the resistance to twisting moments within the femoral canal also requires analysis.

Forces Acting on the Acetabulum

Many more detailed analyses of the biomechanics of the hip have been directed toward the stresses within the femoral stem than within the acetabulum. However, in the long-term follow-up of Charnley, acetabular loosening has been an important problem (1). The intact acetabulum is a horseshoe form that wraps around the superior, anterior, and posterior aspects of the slightly eccentric femoral head. In the lightly loaded state, the dome of the acetabulum is relatively unloaded, and the stress is transferred from the femoral head to the acetabulum through the anterior and posterior extensions of the horseshoe. As the load is progressively applied, since the acetabulum is not in continuity inferiorly, the anterior and posterior sides of the horseshoe are free to expand so that a more congruous seating of the femoral head is allowed. As Radin has pointed out, this phenomenon of deformation under load leads to increasing congruity with progressive loading (8). If the hip were fully congruent in the acetabulum, full loading would produce incongruence as the anterior and posterior extensions of the horseshoe would separate away from the femoral head on loading. This deformation of the acetabulum under load has relevance to total hip arthroplasty since loading of a deformable polyethylene cup could lead the polyethylene to separate from the acetabulum due to the deformability of both materials.

The analysis of the forces acting on the femur also apply to the acetabulum. The orientation of the resultant vector passing through the acetabulum should pass through the center of the body of the ilium (see Figure 1.2). If there is protrusio acetabuli, then this force will pass through the medial wall, which will ultimately fail with progression of the protrusio. If the vector is lateralized or the acetabulum dysplasic, subluxation and lateral acetabular hip erosion may occur.

Vasu, Carter, and Harris have analyzed the distribution of stresses in the acetabulum before and after total hip replacement, using finite element analysis (10). In the normal hip they found transmission of compressive stresses by the cancellous bone of the body of the ilium to the lateral acetabulum wall and lesser order tensile stresses to the medial wall. After conventional total hip replacement, the compressive stresses in the cancelbus bone immediately above the cup were increased, as well as tensile and compressive stresses in the medial wall. Stresses in the lateral wall were decreased. Adding metal backing to the cup redistributed the stresses throughout the whole acetabulum so that stress in the cancellous bone was reduced.

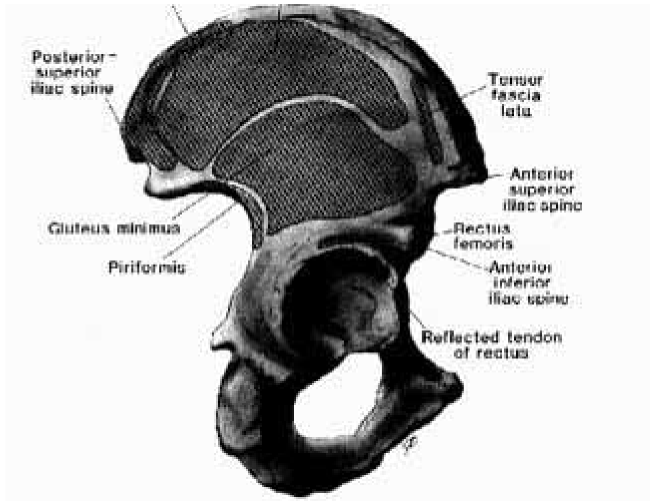

***Figure 1.29*.** Lateral view of pelvis and principal muscle origins.

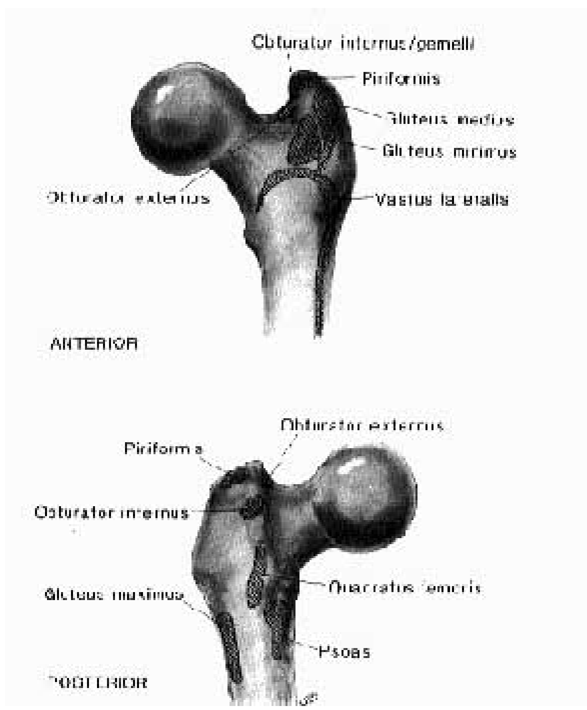

Figure 1.30. Anterior and posterior view of the proximal femur, showing principal muscle insertions.