BIOMECHANICAL PRINCIPLES OF THE HUMAN HIP JOINT

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ABSTRACT. Total hip replacement is the most commonly practiced orthopedic surgery of our days. But to perform this surgery with best results, we have to understand, how the coxofemoral joint work, we must understand it’s biomechanics. To explain the forces acting on the hip joint, and to explain the behavior of the proximal end of the femur, a series of experimental studies were made, that used very sophisticated equipment. The progress of modern imaging (especially acquisitions of computer tomography of the hip), appearance of tridimensional extensometer transducers, hip simulators, mathematical tool improvements (especially Finite Element Method) and software techniques, have allowed the development of complex spatial patterns, modeled after the features of the human hip. Basic knowledge of biomechanics is essential for understanding the hip joint, and after that to understand which disturbance of normal biomechanics can lead to arthritic lesions.

KEYWORDS: BIOMECHANICAL PRINCIPLES, HIP JOINT, HIP REPLACEMENT.

INTRODUCTION

Total hip replacement is the most commonly practiced orthopedic surgery of our days. But to perform this surgery with best results, we have to understand, how the coxofemoral joint work, we must understand it’s biomechanics.

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The first who tried to explain the forces acting on femur was Künstcher in 1934, who used varnish and enamel adering to the surface of the femur, to highlight cracks appeared at that level during different mechanical tests. Later Milch, in 1940, Pauwels in 1954 and Blaimond in 1963 studied the same forces with photoelasticity, but the contemporary progresses allowed to be made more accurate and relevant appreciations of the hip joint, under various biomechanical stress.

As a starting point they used Fischer’s and Braune’s reserches of human walking and they could established the aproximate muscle forces acting on the hip joint during various activities of daily living.

The progress of modern imaging (especially acquisitions of computer tomography of the hip), appearance of tridimensional extensometer transducers, hip simulators, mathematical tool improvements (especially Finite Element Method) and software techniques, have allowed the development of complex spatial patterns, modeled after the features of the human hip. The enormous amount of new informations, made over the last decades, have allowed approximating the complex behavior of this joint, but we wasn’t able to elucidate all physiological and pathological aspects of this joint.

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JOINT GEOMETRY AND FORCES ACTING ON THE HIP

From theoretical perspective, coxofemoral joint can be considered a spherical joint, the femoral head and acetabular cup made of. In fact, the congruence of the two joint surfaces is not perfect, but is improved by the presence of articular cartilage, which has a variable thickness from one area to another (Goodfellow 1973).

During the joint movement the two systems are rotated relative one to another about a common center of rotation, wich is their geometric center. This is a virtual point which carries the forces when we acting the hip. Most of these biomechanical forces are forces of pressure, which causes tension.
at the joint, but these tensions are unevenly taken from different areas of the articular surfaces of contact. Trueta (1954) was the first author who introduced the concepts of bearing surfaces and surface discharge. Under these conditions the two surfaces can be characterized by two coordinates: latitude and longitude (similar to terrestrial globe), the pole capital is considered the center of femoral fossa and the ecliptic pole is the center of acetabulum.

We also have bearing surfaces (pressure or support), which are the articular contact areas. Here the permanent mechanical stress exercises his force when we are acting on the hip. These contact areas are located symmetrically but mirrored on the femoral head and on the acetabulum. This area on femoral head is located in the upper portion of the pole having an average surface area of about 12 cm².

The discharge surfaces means articular contact areas that are only occasionally under pressure. At the femoral head are described two classic discharge surfaces, one located at the lower inner joint portion of the head (under the round ligament), and the other located on the external side of the femoral head (corresponding the acetabular roof).

Important is that the bearing surface of the femoral head is smaller than the articular contact surface, this in turn is smaller than the anatomical surface. Also, the size of bearing area depends of acetabular surface which can be correlated with the size of the coverage of the femoral head angle (the angle of Wiberg).

Modern studies highlights the fact that in reality the human hip joint is incongruent, acetabulum having a smaller diameter than the femoral head and in biomechanical loading conditions the acetabulum suffers an elastic deformation, becoming congruent with the femoral head. (Goodfellow, Greenwald, Afoke, 1986).

**PHYSICAL AND MATHEMATICAL MODELS OF THE HUMAN HIP**

Over time, many authors attempted to determine the different measurements and using a range of physical and mathematical models, the size of the task that supports the hip in various stress conditions. Reference is Fischer's research (1895) that determined the size and direction of the forces acting on the coxofemoral joint in every phase of gait (See Figure 1).

![Fig. 1 Fisher’s Phases of Gait](image)

Pauwels starting from studies of Fischer calculated the direction of the forces acting on the femoral head in frontal plane. He determined that the support forces acting on the hip have only vertical direction and in unipodal support the center of gravity of the body is projected from about 10.9 cm from the center of the femoral head. So standing with symmetrical bipod support, the center of gravity of the head, trunk and upper limbs is on the mediator line joining the centers of the femoral heads.

The whole study concludes that the magnitude of the resultant forces acting on coxofemoral joint varies in intensity during gait phases. It reaches the maximum value (equal to five times the weight of the body) at the gait position where the heel contacts the ground, toe leaves the ground, respectively, and decreases to a minimum value (equal to four times the weight of the body) in position when the whole plant is sustained on the ground.

**Variations of forces acting on the hip during walking**

Variations of the forces presented above took into account only static forces acting on the hip. In fact, during walking, the center of gravity ($S_z$) oscillations appear in all three planes of space, in addition, body mass acceleration trigger additional inertial forces (D). As a result, the weight of the moving body, becomes the sum, between the
weight of gravity of the body and the vertical component of the force of inertia.

The tension force “R”, acting on the femoral head in vertical direction, has a varying amplitude, reaching the maximum value, equal to about 4.4 times of body weight (a weight of 2580 N to 587 N) when the lower limb becomes carrying.

The horizontal component of the force of inertia (D₀) was deduced by Pauwels from the horizontal acceleration of the center of gravity S₅. This component tends to rotate the pelvis around the femoral head in clockwise, being offset by muscular contraction M₀ caused by external rotator muscles (piriformis, gemellus superior, obturator internus, gemellus inferior, obturator externus, quadratus femoris).

The resultant forces R₀ acting horizontally on the femoral head while walking was assessed by Pauwels as having a maximum value of 432 N for a person weighing 587 N (for a normal walking speed in forward direction).

The center of gravity of the body S₅, is being located higher than the center of rotation (O) of the head of the femur, so the horizontal component of inertia force in this case will result one phase in frontal plane and one in sagittal plane, but their value is not significant. By calculations, Pauwels showed that forces requests stronger the femoral head in the frontal plane than in the horizontal plane.

The resultant of the forces (R), acting on the femoral head causes a reaction force (R₁) which tends to modify the position of the femoral head in the acetabular cup by two components: a longitudinal one (P) with vertical direction which tends to displace upwards the head of the femur, and other transverse one (Q), parallel to the bearing surface, with much lower value compared to P, which tends to push the femoral head to the acetabular floor.

The resultant R changes direction when walking resulting from displacement of the center of gravity of the body.

Taking as reference the three characteristic phases of unipodal support (after Fischer and Braune: phases 12, 16 and 22), Bombelli (1983) indicated that the resultant R is antero-latero-caudal in phase 12, latero-caudal in phase 16, postero-

![Fig. 2 Variations of direction of the resultant R and Q component during walking](image)
These purely mechanical data only approximates the behavior of hip joint because the quantitative comparison of each muscle and temporal forces of their actions is almost impossible to assess. This explains the significant differences between results obtained by different researchers who described theoretical muscle forces during walking. In addition, the forces performed by periarticular ligaments and soft parts are very difficult to calculate, because it was not done so far an anatomical model of "passive elements" of hip. However, some studies have been conducted to assess the need for moving of a joint, while the periarticular musculature is relaxed (so called passive resistance of the joint). It has been shown that the joint’s passive resistance is visco-extensible which is generated by the deformation of periarticular soft tissues and ligaments.

The attempts of quantitative evaluation of the resistance during a normal gait cycle, showed that its value remains less than 10% of the exterior moment and greatly increase when movements of flexion more than 40° and extension more than 105° of the hip are made (Yoon and Mansour, 1982).

Many authors consider as a reference system for assessing global force R acting on the hip joint which resulted in values of R according to body weight G for different particular situations, as follows:

- at rest and bipod support: \( R = \frac{1}{3} G \);
- at rest and unipod support: \( R = 2.5 – 3 G \);
- while walking: \( R = 4 \) to 4.5 G;
- when climbing a ladder: \( R = 6 – 8 G \);
- unipod support with a crutch in the contralateral hand: \( R = 0.8-1.2 G \);
- unipod support with a crutch in the homolateral hand: \( R = 1.5-2.5 G \);
- patient lying, hip flexed and knee extended: \( R = 1.5 G \);
- patient lying, hip flexed and knee flexed: \( R = 0.9 G \);
- patient sitting in the chair: \( R \) very weak.

These results are obtained by using an Austin-Moore prosthesis (with pressure transducers placed percutaneously), while walking on an electronic treadmill but this results are contested today by some researchers who consider that determination methods are outdated.

DISTRIBUTION OF PRESSURES AND TENSIONS INDUCED IN THE HIP

The forces acting on the hip are inducing varying tensions of the articular contact surfaces. Photoelastic studies of H. Fessler (1957) showed that in the hip, the tension trajectories of photoelastic models are similar to those of trabecular bones, while the author has shown, that the biggest changes in tension distribution of the acetabulum are determined by pelvic tilting movements against the femur.

Dietsch et al. (1974) were able to determine experimentally that the compressive tensions inside the acetabulum reach normal values of 0.8 Newton / mm². Experimental studies performed in vitro, indicates that acetabular dome supports very high pressure: approx. 2 megapascals (2 mega Newton / m²), applied with a cyclicity of 90,000 cycles (Greenwald, 1972, Freeman, 1973 Hodge, 1989).

Measurements in vivo (using instrumented prostheses), shows that during different phases of
gait, the pressure on the acetabular dome can reach 5.5 megapascals, and a normal hip is requested to support more than 1 million annual cycles, also the acetabular pressures greatly increases when daily activities are done such as climbing and descending stairs, rising from a chair, jumping from one level to another (see Fig. 4).

When the hip's biomechanical demands are weak (16% of body weight), the upper pole of the femoral head does not touch the cotyloid dome, articular contact is “peripheral para equatorial type” (see Fig. 4).

The progressive increase of the load of the hip to a value of approx. 50% of body weight, forces the femoral head to make contact with the acetabular dome, without losing peripheral contact, achieving complete articular congruence. Greenwald has demonstrated, through a well-developed mathematical model, that during different phases of gait, there are performed infinite “super positions” of the two surfaces and these contacts are forming an “articular gearing”. It works when we are applying high physiological loads, with the role to equalize pressure throughout the bearing surfaces. In conditions of excessive and eccentric loading of the hip (as it happens for example in the case of stair climbing), the polar pressure is becoming more pronounced.

An experimental study in vivo, using instrumented endoprosthesis (with internal pressure transducers) and a sophisticated data monitoring equipment (optoelectronic cameras adapted to a computer) highlighted a particular aspect of intra-articular pressure distribution in the joint. They have recorded high pressures in the dome of the acetabulum (which was known from studies of Greenwald, Bergmann et al) but there was increased pressure and the upper posterior portion of the acetabulum (see figure 4). This has relevance to acetabular dysplasia, where is tempted superio-posterior wall reconstruction with bone grafts, but this increased posterior pressure can shorten graft survival and the hip implant’s life (Hodge, 2009).

A currently controversial subject is the biomechanical role of the acetabular labrum and the transverse ligament, in the distribution of intra-articular pressure in the joint. They have recorded high pressures in the dome of the acetabulum (which was known from studies of Greenwald, Bergmann et al) but there was increased pressure and the upper posterior portion of the acetabulum (see figure 4). This has relevance to acetabular dysplasia, where is tempted superio-posterior wall reconstruction with bone grafts, but this increased posterior pressure can shorten graft survival and the hip implant’s life (Hodge, 2009).

To understand the physiology of the human hip biomechanical study is strongly needed. To perform a more accurate hip replacement the biomechanical values that must initially studied.
Biomechanical forces changes from patient to patient, in different pathologies the values are changed.
Total hip prostheses must meet and reproduce the physiological values of the hip joint. Non-compliance of biomechanical principles soon lead to the destruction of the prosthetic implant.

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REFERENCES


