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Review articles

Evolution of the Koch model of the biomechanics of the hip: clinical perspective

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Abstract Over the past several decades numerous researchers have revisited the model of the biomechanics of the hip first predicated by John Koch in 1917. The contributions of Blount (1956), Frankel (1960), Pauwels (1976), Toridis (1969), Rybicki (1972), Fetto (1994, 1995), Ling (1996), and Lu (1997, 1998) among others created a more complete picture. The present article briefly reviews the previous biomechanical concept and its clinical inconsistencies and offers a model that includes the dynamic and static input of the soft tissues. The action of the iliotibial band (ITB) and the vastus lateralis–gluteus medius complex (as static and dynamic tension bands lateral to the femur) counterbalance the varus bending torque of the loads acting on the hip, transforming the tensile stresses in the lateral femur (as hypothesized by Koch) into compressive stresses. The inclusion of the soft tissues, extending the previous model, widens our understanding of the forces acting on the hip. Thus, a variety of clinical observations can be better explained in a comprehensive theoretical framework.

Key words Hip · Biomechanics · Iliotibial band (ITB) · Lateral flare

Koch's model

The classical work of John C. Koch[14] published in the *American Journal of Anatomy* in 1917 includes a geometrical description of the femur and a calculation of stresses induced by force loadings that were assumed to occur during gait. He believed in the accuracy of the model of proximal femur loading defined by Culmann,[4] which was based on an assumed analogy to the Fairbarn crane. Koch correlated the stress patterns in the trabecular bone with Wolff's concepts of bone formation.[31]

This model was based on cadaveric studies in which a 100-lb force was directly applied in a downward direction to the head of a young man's femur. He calculated the relation of the body and the abductor muscles lever arms to be in a 2:1 ratio, leading him to state that the abductor muscles (gluteus medius primarily) must generate twice the force of the body's weight to maintain equilibrium, preventing the body from falling toward the unsupported side. Based on his study he assigned positive and negative values, representing compressive and tensile forces, respectively, along the medial and lateral femoral surfaces. He specifically noted that the superior neck and proximal lateral three-fourths of the femoral shaft experience tensile loads, whereas the distal lateral femur and the entire medial femoral surface were thought to experience compressive loads (Fig. 1). However, he did not provide any explanation for the conversion from tensile to compressive loading in the distal lateral femur.

Koch's relevance

Koch's model was in agreement with observations concerning fractures of the proximal femur, all of which tend to collapse toward varus, supporting the medial compression–lateral tension hypothesis. It was also consistent with the fact that both femoral condyles experience compressive loads, albeit unequal, in the medial and lateral knee compartments. His treatise was so powerfully presented it stood unchallenged as the definitive model of hip biomechanics for the next 70 years. It also served as a foundation for the design, testing, and validation of hip replacement prostheses.

Deconstructing Koch

Koch's work was bounded by the technological limitations of his era and could not measure other variables acting on the hip joint during gait. As a matter of fact, Koch himself acknowledged this limitation. He dis-

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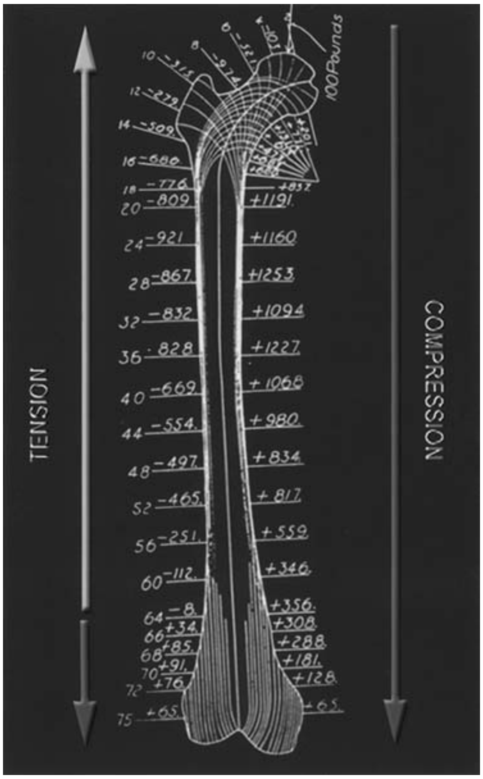

Fig. 1. Koch's model of the hip (based on an assumed analogy with a cantilevered crane). Under his theory there is lateral tension and medial compression along the femoral cortices

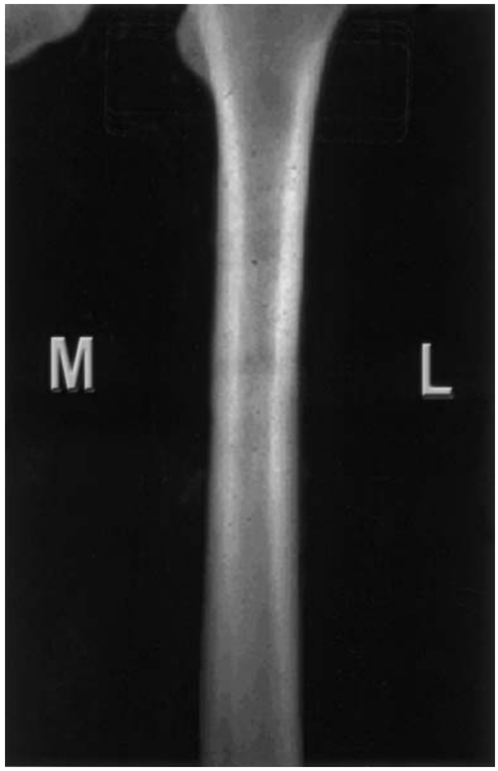

Fig. 2. Radiograph of a normal femoral shaft. Note the quality and quantity of cortical bone along the lateral and medial surfaces. The lateral cortical bone stock approximates 75%–80% of that of the medial cortex. *M*, medial; *L*, lateral

missed the effect of the muscular activity on the loaded femur as being "relatively small and very difficult to analyze." As presented, his model was based on the effect produced by an isolated vertical weight acting on the femoral head. As a static model, it did not consider the function of the soft tissues around the joint and did not provide answers to important questions regarding bone morphology, osteology, and energy expenditure.

Adaptive remodeling of the bone

According to Wolff's law,[31] bone is formed in response to the quality and quantity of the load it experiences. If there is compression at the medial and lateral compartments of the knee and if there is (as Koch stated) medial compression and lateral tension at the proximal femur, the model fails to provide an explanation as to *where* along the lateral cortex the *transition* from tensile to compressive forces occurs. Such transition implies a change in the mechanical environment and, according to Wolff's law, a subsequent morphologic change in the loaded bone. The normal human femur does not demonstrate a specific site for this transition from tensile to compressive load. There is, in fact, continuity of cortical bone along the lateral femur, from the apophysis of the greater trochanter, increasing in thickness throughout the diaphysis and diminishing in mass at the distal femur, to finally disappear at the level of the lateral femoral epiphysis (Fig. 2).

Femoral neck–shaft angle dilemma

At birth, the femoral neck–shaft angle is greater than 160°. As bipedal gait is initiated, the neck–shaft angle decreases to a final angle of 130° *long before skeletal maturity is reached*. The varus torque at the femoral neck level continues throughout our bipedal existence. During childhood the uncalcified growth plate is capable of reacting and remodeling in response to a mechanical stimulus. Despite this flexibility, the neck–shaft angle reaches a steady value before maturation of the epiphyseal plate (Fig. 3).

A related factual observation has to do with surgical correction of an excessive valgus neck–shaft angle (varus osteotomy): if there is additional growth potential in the femoral epiphysis, the neck–shaft angle increases toward valgus from the varus position created at the time of surgery.[20] Koch's static model of varus loading does not account for any other counterbalancing force that helps explain why this occurs or why the neck

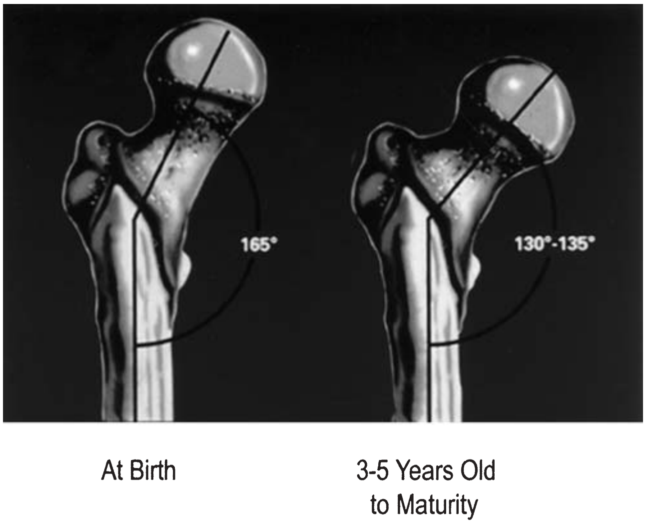

Fig. 3. Neck–shaft angle progression. From 165° at birth to the final value of 130°–135° at 3–5 years of age (well before maturation of the growth plate)

shaft angle stops at approximately 130° before skeletal maturity is reached.

Spending energy

Koch correctly noted that during the unilateral support phase of the gait there is an inward varus stress across the femur. It was widely accepted that this varus torque was counterbalanced by the action of the gluteus medius.[2,3,9,12] However, the amount of energy to sustain this biomechanical effort during an extended period of walking exceeds the metabolic capacity of the gluteus medius. This problem in energy expenditure is prominent in amputees. Below-knee amputees experience only a small increase in energy expenditure (10%) compared to the intact individual; however, with the above-knee amputee there is a significant increase in muscular demand to maintain an equilibrium state during unilateral stance (40%).[30]

When we consider the surgical level between a below-knee amputee and an above-knee amputee, we find that the only static stabilizer of the pelvis being violated when transecting the limb proximal to the knee joint is the iliotibial band. Moreover, below the knee amputation (BKA) patients, because of their intact gluteus medius muscles, walk without a noticeable limp and can stand without listing toward the amputated side (a negative Trendelenberg sign). Despite having intact gluteus medius muscles, above the knee amputation (AKA) patients can only stand listing toward the amputated side (a positive Trendelenberg sign) and walk with a distinct limp.

One of the mainstream arguments when explaining the difference in energy expenditure in amputees relies on the absence of the knee joint mechanism. Knee flexion in stance is the third determinant of gait described by Inman[12] to minimize the path of the center of gravity during ambulation and thus minimize the energy expenditure. The deficit of this compensatory mechanism increases the rise of the center of gravity by 7/16 inch.[5] The destabilization of the pelvis that takes place after the transection of the iliotobial band (ITB) during AKAs[8] affects directly the sixth determinant of gait or lateral displacement of the pelvis. The ITB insufficiency cannot compensate for the resulting mediolateral pelvic sway (1.7 inches). The dramatic difference in energy expenditure between AKA and BKA patients is better explained when these factors are considered together.

Trendelenberg sign: gluteus medius insufficiency?

As aforementioned, it was argued that the main factor accounting for the variation in energy expenditure was the loss of the tensional effect on the gluteus medius provided by the adductors. This in turn creates an "apparent gluteus medius weakness," explaining the positive Trendelenberg sign associated with AKAs and its absence in BKA patients.

Anatomically, the adductor longus inserts distally via an aponeurosis into the medial lip of the linea aspera of the femur. The adductor magnus fans out to take distal insertion in continuity to the gluteal tuberosity and via a broad aponeurosis to the linea aspera (deep to adductors brevis and longus) and the medial supracondylar line down to and including the adductor tubercle. Assuming that complete section of the adductors is a surgically sound option when performing an AKA, we still cannot explain the fact that all AKA patients, regardless of the transection level in the thigh (i.e., high AKA or low AKA) exhibit a significant limp. Theoretically, in low AKA patients the intact adductors should exert significant "tensional" compensation for the gluteus medius to perform efficiently and thus to avert a Trendelenberg sign. The only static stabilizer of the pelvis being transected in all AKAs is the ITB. As stated before, the ITB compensates for a sway in the mediolateral plane of 1.7 inches during gait and, when transected, for a mediolateral inward sway of more than 30° in cadaveric experiments.[7]

Function and dimensions of the gluteal muscles

Koch predicated that the gluteus medius must exert a force equivalent to two body weights to equilibrate the pelvis. Several investigators have shown that the 2:1 ratio is a conservative estimate and that the actual force that must be counteracted during midstance exceeds three body weights.[15,17] According to electromyographic studies, the gluteus medius is most active just before midstance.[6,12] Its activity then decreases, being significantly less active during the actual midstance phase of the gait. This contradicts the prediction of Koch's model where the gluteus medius should be under the greatest demand *during* midstance. Even though the gluteus medius has to generate a herculean force to reach mechanical equilibrium across the hip joint, its rather small size does not seem to reflect its important function.

This size–function divergence becomes more prominent when considering the size and strength of the most robust muscle in the human economy: the gluteus maximus. Only the inferior portion of the muscle finds insertion in the gluteal tuberosity. Most (75%) of its fibers (along with the tensor fasciae latae) converge into the ITB. The dynamic action of these muscles and the static input of the ITB throughout the mid and late stance phases of the gait maintains pelvic stability and acts as a lateral tension band along the femur.

Fractures and tensile stresses

Frankel,[9] Inman,[12] and others[2,3] have suggested that contraction of the gluteus medius provides protection to the femoral neck against fractures. They have stated that the varus load experienced by the femoral neck during the unilateral stance phase would otherwise cause the superior neck to fracture more readily than what is usually observed. Rybicki et al.[23] demonstrated that the amount of tensile loading the gluteus medius must create in the proximal lateral femoral diaphysis to protect the femoral neck against the varus torque would be unphysiologically excessive. They calculated that during unilateral stance these loads would be nearly 70% of the maximum tensile stress sustainable by the intact femur.

The reduction and fixation of subtrochanteric fractures is a common procedure in orthopedic practice. Although several methods have been used in the past, the "sliding compression screw" originally described by Allgower et al.[1] is widely regarded as the optimal treatment for intertrochanteric fractures of the femur, permitting bone fragments to impact until a bony support has been established across the fracture site. The plate must be placed on the lateral side of the femur (the tension side of the classic model). If the lateral femoral cortex is physiologically loaded in tension, the lag screw and the plate experience the type of mechanical stress (tensile) that would never allow stabilization of the fragments, let alone their consolidation and healing, because *fractures simply cannot heal under tension* (Fig. 4).

When compared with other fixation modalities, compression hip screws are three times stronger than the intramedullary Ender's nails in simulated unstable intertrochanteric fractures.[11] This fact cannot be sustained concomitantly with the existence of a lateral cortex loaded in tension.

Fig. 4. Radiograph of a dynamic compression hip screw after a complex femur fracture. Note the compression plate on the lateral side of the femur and the sliding screw backing out under compressive forces (*arrow*)

Beyond Koch: dynamic model of the hip

The ITB, also known as the band of Maissiat, has been the subject of several investigations. It drew the attention of Jacques Maissiat[19] in 1843, who considered it to be primarily a ligament connecting the ilium with the knee and the principal factor maintaining the balance of the body in stance and motion. Thomsen[25] in 1934 looked at the matter even closer and concluded that the function of the ITB to limit adduction enables patients with hip paralysis to stand and even walk.

Hermann Meyer (1853) also believed that the ITB, through contraction of the vastus lateralis muscle, could be tensioned by the resulting force acting along the direction of the band. This would keep the pelvis in equilibrium during one-legged stance, thereby decreasing the workload of the abductors. Pauwels[22] demonstrated that tensioning of the ITB during one-legged stance would decrease the bending stresses on the proximal femur and correspondingly increase its axial loading.

Jacob and Huggler[13] investigated the function of the ITB in vivo and recorded its degree of lengthening and tensioning during gait. They concluded that the varus bending stresses in the femur could be partially relieved by tensioning the ITB up to one body weight, supporting the 1975 view of Oberlander[21] that absence of the tract might develop a fatigue crack below the greater trochanter.

Rybicki et al.[23] analyzed mathematically the stress in the femur using three models: (1) the simple axial loading model (Koch); (2) the axial loading model with inclusion of the abductor muscle forces; and (3) in the final phase of the study, addition of the ITB and tensor fascia latae force. They observed that by increasing the force exerted by the ITB the pattern observed in the simple axial loading model (Koch) of medial compression/lateral tension was reversed. It was concluded that with in vivo conditions the ITB would produce an appreciable reduction of stress: 45% for 135 lb of tension in a 200-lb man. It has been stated that "if the human body is considered to be an efficient mechanical structure . . . then the tensile stresses in the femur should be much lower than the compressive stresses, since bone is about twice as strong in compression as in tension. This condition is met with the 200-lb tension in the tensor fascia latae, but not in its absence or with excessive tension of this muscle."

Fig. 5. Trabecular blueprint of the femoral neck and head. As stated by Garden and St. Clair Strange, both superior and inferior systems (*arrows*) transmit the compressive stresses into the femoral cortices

Revisiting the tension band hypothesis

The concept of a tension band (AO group) is that of an inelastic band that when placed under tension creates a compressive load medial to that band proportional to the medial displacement of a point from that band. In 1994 and 1995, one of us (J.F.)[7,8] proposed a model that included the ITB as a static lateral tension band. The importance of the soft tissues as tension elements relieving the varus bending torque of the femur was subsequently reaffirmed and described by others.[16–18]

There is a minimal amount of cortical bone at the distal aspect of the greater trochanter. As we move distally along the femur from the level of the trochanter toward a level 2 cm distal to the inferior border of the lesser trochanter (where the ITB is at its most distant point from the lateral cortex), the mass of cortical bone in the lateral side gradually increases until it reaches a maximum value of approximately 75% of that of the medial cortical bone. This lateral cortical bone mass remains constant until the distal third of the femur, where it diminishes and eventually disappears upon reaching the level of the distal lateral epiphysis (where the ITB is closer the lateral cortex).[8]

In 1892 Wolff offered "the law of bone remodeling."[31] He analyzed the particular disposition of the various groups of trabeculae in the proximal femur and likened their two-dimensional arrangement to a Fairbairn crane, adopting views similar to those of Culmann[4] and Ward.[29] He demonstrated that the cortical bone is a highly dense layer composed of distinct groups of trabeculae and that it owes its particular morphology to the magnitude of the load acting on it. What is known today as Wolff's law implies that the bone is responsive to its mechanical environment and that the loads applied have a structural correlation with its morphology.

At the proximal femoral epiphyseal level, the complex trabecular pattern compensates for the intrinsic inadequacy of the subchondral (cancellous) bone to resist the compression load generated at the articular surface. This intricate trabecular blueprint (Fig. 5) is responsible for conducting compressive stresses to the diaphyseal cortices, as argued by St. Clair Strange[24] and Garden.[10] As body weight increases the load in a downward direction, the thin layers of trabecular bone conform themselves along the direction of the main loads (as described by Wolff) into a denser, more compact structure. Thus, the appearance of increasing cortical bone along the femur represents the physiological adaptive response of the bone to the gradual change in the mechanical environment.

The mathematical shortcomings of the static model of the biomechanics of the hip outlined by Rybicki et al.[23] and the irreconcilable differences between the Koch model and the clinical observations outlined above (i.e., the presence of cortical bone in the lateral "tension" side of the femur, the energy expenditure and Trendelenberg dilemma in amputees, and the natural progression of the neck–shaft angle) served as the theoretical and applied-science foundation of a dynamic

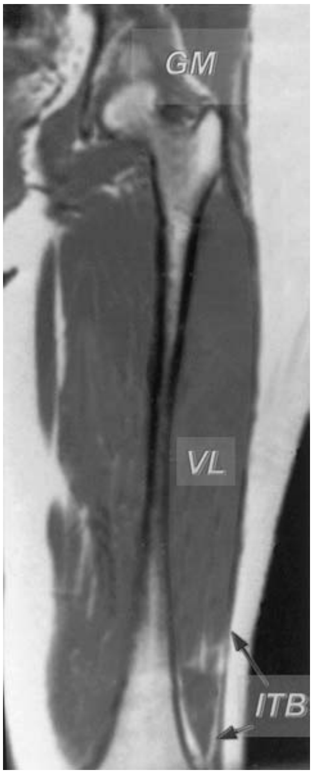

Fig. 6. Magnetic resonance (MR) image of the thigh in the coronal plane. Note the anatomic importance of the soft tissues around the femur. *ITB*, iliotibial band; *GM*, gluteus medius; *VL*, vastus lateralis

Fig. 7. Anteroposterior radiograph of a lateral flare stem designed to engage only the metaphyseal femur. The lateral flared design permits use of the lateral cortex as an additional base of support for the prosthesis

model encompassing the ITB as a static (ligamentous) tension band and the vastus lateralis–gluteus medius complex as a dynamic one along the lateral femur moderating the bending moments (Fig. 6).

Static input

The importance of the ITB as a significant static stabilizer of the pelvis during unilateral stance was demonstrated in cadaveric studies by Fetto and Austin.[7] It was found that, in the presence of an *intact* ITB, sectioning the capsule, gluteus minimus, gluteus medius, and gluteus maximus permitted the pelvis to rotate inwardly a mean 10° from the initial horizontal position about the hip joint. However, in the presence of an intact capsule, gluteus minimus, gluteus medius, and gluteus maximus, but a *sectioned* ITB, there was a mean 30° of inward rotation of the pelvis about the femoral head.

Dynamic input

The greater trochanter represents the insertion place of the gluteus medius and the vastus lateralis. It creates the same type of mechanical advantage (in the coronal plane) as the one produced by the presence of the patella between the quadriceps and the patellar tendon (in the sagital plane).

The contribution of these muscle forces as dynamic tension bands along the lower extremity has been more recently confirmed in in vivo studies. Through the use of gait analysis and telemetry of massive femoral components implanted in two bone tumor patients, Lu et al.[17,18] showed the effects of muscle activity on the axial compressive strain on the femoral diaphysis. They concluded that the "bulk of the bending moment along limbs is transmitted by a combination of tensile forces in muscles and compressive forces in bones, so moments transmitted by the bones are smaller than the limb moments."

Conclusions

As a result of inclusion of the ITB and the gluteus medius–vastus lateralis complex, there is no tension loading but, rather, compression loading throughout the lateral femur from the greater trochanteric area to the distal lateral epiphysis during unilateral support phase of gait. The creation of a "compression gradient" between medial and lateral cortices instead of the previously described lateral/tension and medial/compression pattern depicts physiologically the loading forces acting on the hip joint and along the femoral shaft. This is consistent with bone morphology, the changes of the neck–shaft angle during growth, clinical observations of energy expenditure, and in vitro recreations of the human condition.

Implications

The ITB model of hip loading has a significant direct impact on prosthetic materials and design considerations when developing femoral components for total joint replacement. This model demonstrates the necessity of a total hip replacement femoral component to engage the proximal lateral femoral cortex as an additional area of support against subsidence. Specifically, this model dictates that this engagement should be at or proximal to the intersection of the central neck axis and the lateral femoral cortex to avoid stress shielding and hence loss of proximal femoral bone (Gruen zones I and VII). This geometry prevents nonphysiological diaphyseal loading, which may be a significant cause of microfractures, thigh pain, and consequent diaphyseal hypertrophy after total hip replacement[26–28] (Fig. 7).

Rehabilitating Koch

Koch's work was circumscribed by the technological limitations of his era. Nevertheless, he had a clearer vision than the one sometimes ascribed to him. In 1917 he acknowledged that "the chief function of the bones from the mechanical point of view is to serve as supporting structures, the stresses in bone being chiefly compressive."[14]

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