(i) Basics of orthopaedic biomechanics

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Abstract

An outline of the basic principles of orthopaedic biomechanics is presented. Joint moments, muscle moment arms, *in vivo* forces, contact stresses and joint stability are all discussed with recent clinical examples to demonstrate their importance. These clinical examples focus on the hip and the knee and include: the effects of femoral offset and reducing the abductor moment arm on hip arthroplasty, how the knee adduction moment causes an asymmetric load distribution between the condyles, the magnitude of *in vivo* forces and their implications for wear, the consequences of meniscectomy on cartilage contact stresses, extreme contact stresses caused by edge loading in hip replacements, the effect of femoral head size and capsular repair in total hip replacement stability, knee medial rotation and the role of the anterior cruciate ligament in joint stability.

Keywords biomechanics; contact stress; hip; knee; stability

Introduction

The forces and moments acting about a joint control how it moves and functions. It is important to understand these loads during activities, because abnormal loading may lead to joint degeneration and osteoarthritis.[1,2] Moreover, the clinical problems following joint replacement, which is the principal way to treat severe osteoarthritis, are also largely caused by abnormal loading in the joints.[3,4] Therefore, it is important to have a solid understanding of the basics of orthopaedic biomechanics, including joint torques and muscle moment arms, the forces experienced by the joints *in vivo*, the contact stresses on the articulating surfaces and how the joints maintain stability. The following sections in this review paper address each of these items in turn, including recent clinical examples of how these core principles influence the function of joints.

Joint moments

Understanding moments and moment arms (also known as lever arms) is essential to grasping the basics of orthopaedic biomechanics. Just as applying a force on an unconstrained object creates a linear movement, a moment creates a rotation of an unconstrained lever about a pivot.

Definition:

$$\text{Moment} = \text{Force} \times \text{Moment Arm}$$

A moment arm is defined as the perpendicular distance of the line of action of a force from a pivot.

Hip abductor moment arms example calculation

The hip joint abductor muscles are essential to allow normal gait, and weak abductors (particularly a weak gluteus medius, which is the largest of the hip abductors) can lead to a Trendelenburg gait (a form of limp).

Static analysis: Figure 1a shows a simplified example of a patient balancing their bodyweight on the right leg through muscle activity; the gluteus medius is contracting and creating a force that stabilizes the hip joint. If the patient is stationary, every force must have an equal and opposite reaction force and the moments of bodyweight and the muscle force must be balanced (otherwise motion would occur). When the left foot is lifted, the bodyweight force tends to rotate the pelvis clockwise around the right hip; this is resisted by the hip abductor tension creating an equal and opposite moment. The downwards pull of weight and the muscle force are resisted by the supporting lower limb and thus the femur creates an equal and opposite joint reaction force acting upwards into the acetabulum. The muscle force and joint reaction force can be calculated by balancing moments and forces about the joint. Figure 1b shows a representation of the hip joint and the forces acting on it.

$$\begin{aligned}&\text{Balancing Moments:}\\ &\qquad \text{Anticlockwise Moment} = \text{Clockwise Moment}\end{aligned}$$

$$F_{\text{GMed}}\mathbf{MA} = 0.1 \times 600$$

$$F_{\text{GMed}} = \frac{60}{\mathbf{MA}}$$

$$\text{If MA} = 0.1\ \mathbf{m}: \quad F_{\text{GMed}} = 600\ \mathbf{N}$$

$$\text{If MA} = 0.02\ \mathbf{m}: \quad F_{\text{GMed}} = 3000\ \mathbf{N}$$

It can be seen that the muscle force required to support the upper body depends greatly on the size of the muscle moment arm. In general, most muscles are attached close to the joint and so they have a small moment arm, which requires a large muscle force in comparison to the weight of the torso. The example will continue with a realistic muscle moment arm of 50 mm, giving an abductor muscle force of $F_{\text{GMed}} = \mathbf{1200\ N}$.

The muscle force acts at an angle θ to the vertical. Before calculating the joint reaction force it helps to split this muscle force into two components using trigonometry, one horizontal and one vertical. Taking $\theta = 20°$.

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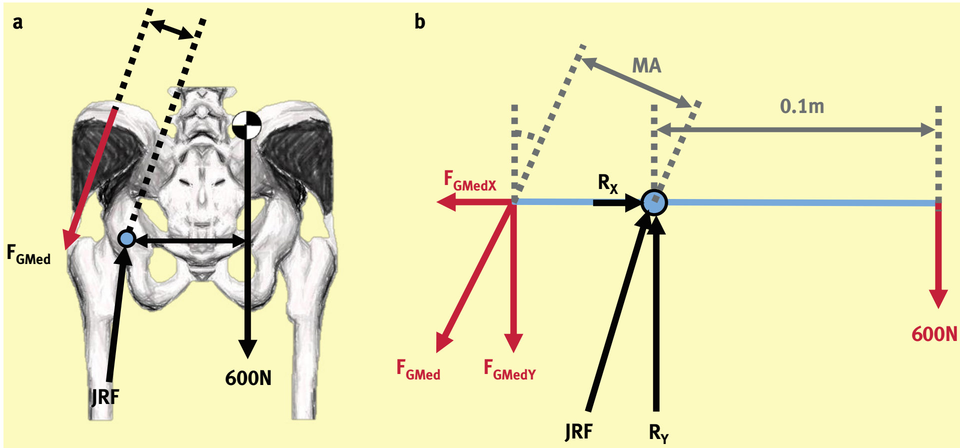

Figure 1 A patient standing on one leg. (**a**) The gluteus medius provides a balancing force (F_{GMed}) for a patient weighing 600 N. The hip joint can be considered a pivot point, and the joint reaction force (JRF) can be seen. (**b**) A diagram of the gluteus medius stabilizing the hip joint. The X and Y components of the joint reaction force (R_X and R_Y) and the gluteus medius muscle force (F_{GMedX} and F_{GMedY}) can also be seen. MA is the gluteus medius moment arm, and θ is the angle the muscle force makes with the horizontal beam.

Resolving forces:

$$\text{Horizontal } (x-\text{direction}):\ F_{\text{GMed}X} = F_{\text{GMed}} \times \sin\theta = 1200 \times \sin 20 = 410\,\text{N}$$

$$\text{Vertical}(y-\text{direction}):\ F_{\text{GMed}Y} = F_{\text{GMed}} \times \cos\theta = 1200 \times \cos 20 = 1128\,\text{N}$$

The components of the hip joint reaction force acting in the X and Y directions can be found by summing the horizontal and vertical forces.

Balancing forces:

$$\text{Horizontal}:\ R_X = F_{\text{GMed}X} = 410\,\text{N}$$

$$\text{Vertical}:\ R_Y = F_{\text{GMed}Y} + 600 = 1128 + 600 = 1728\,\text{N}$$

The joint reaction force is the resultant force and can be found using Pythagoras's theorem:

$$\text{JRF} = \sqrt{R_X^2 + R_Y^2} = \sqrt{410^2 + 1728^2} = 1776\,\text{N}$$

It can be seen that for this two dimensional example, the reaction force at the hip is almost three times the weight of the torso. *In vivo*, the forces and moments must be balanced in all three dimensions and muscle agonist and antagonist co-contraction can enhance joint stability. An example of this co-contraction might be the addition of rectus femoris tension, which is an antagonist to the hamstring muscles during hip joint extension, adding additional force to the hip joint. The result is that *in vivo* joint reaction forces are high: ranging from 2 to 3 BW (bodyweight) during gait, and up to 10 BW when stumbling.[5,6]

Dynamic analysis: joint motion occurs when the joint moments are not balanced. The difference between the clockwise and anticlockwise moments results in an accelerating moment (or torque), which causes the joint to rotate. The analysis method for a dynamic situation is much the same as the static analysis, with the addition of a torque variable that is proportional to the joint acceleration. This analysis requires knowledge of additional factors such as the inertial properties of the limb, and is necessary if an accurate analysis of complex function such as gait is required.

Femoral offset and the abductor moment arm

The abductor moment arm is of particular relevance for total hip arthroplasty surgeons, where the femoral head and neck are excised and replaced with an implant. The size and varus/valgus positioning of the implant directly affect the femoral offset and hence the abductor muscles' moment arms (Figure 2). Patients with insufficient abductor moment arms require a bigger muscle force to attain the same function during gait. This higher force has two negative effects: first, the gait cycle becomes less efficient because the muscles contract harder and hence use more energy. Second, it increases the joint reaction force and hence contact pressures in the joint, leading to higher wear rates.[7] It is also likely that the patient will develop a limp if their abductor muscles are not strong enough to provide the additional force required to support the bodyweight adduction moment.[8] A reduced femoral offset can also result in soft tissue laxity that may allow a small subluxation of the femoral head during gait. This microseparation can lead to increased wear rates and edge loading of implants, and can lead to prosthetic loosening.[9] However, the femoral offset should not be increased past the design point of the prosthesis used, as an increased offset places a greater bending moment on the implant itself, which can lead to fractures of the prosthesis.[4] The original Charnley hip had a large offset, in order to reduce the joint force and wear rate, but this did indeed suffer from fatigue fractures of the femoral stem.

Knee adduction moment

During gait, the centre of mass is medial to the contact of the foot on the ground, and thus the line of action of the ground reaction force passes medially to the knee joint (Figure 3). The effect of this is an asymmetrical loading profile in the knee, with the

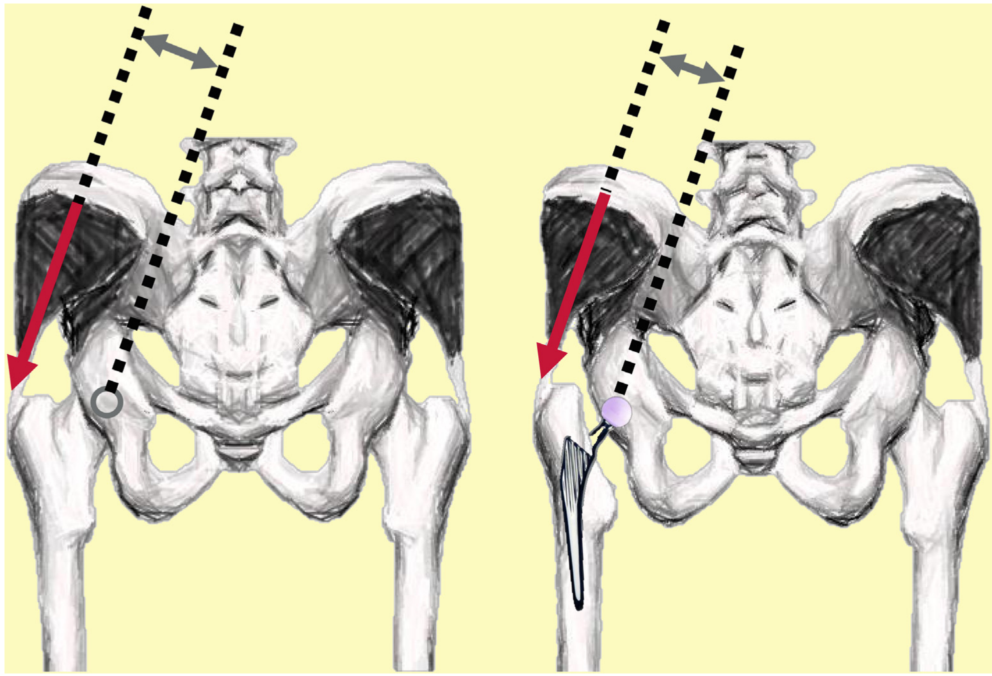

Figure 2 The effects of changing femoral offset on the abductor moment arm. Left, a natural hip joint showing the moment arm of the gluteus medius. Right, a reduced femoral offset due to an implant with a short femoral neck and valgus positioning decreases the abductor moment arm.

medial compartment taking a greater share of the load.[10] In a normal knee, the adduction moment is resisted actively by the quadriceps, gastrocnemius and iliotibial band tension, and passively by the ligaments of posterolateral corner of the knee.[11]

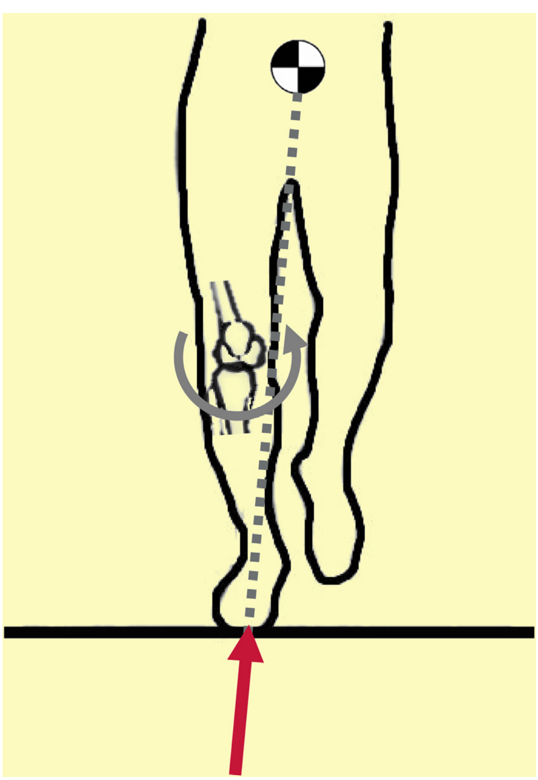

Figure 3 The line of action from the centre of mass to the foot passes medial to the knee, producing an adduction moment during gait.

Patients developing osteoarthritis have been shown to have higher knee adduction moments[1] and hence higher loading in the medial compartment of the knee. Varus knee deformity results in the knee joint being further away from the line of action of the centre of mass and hence there is a bigger knee adduction moment arm and hence higher medial knee loads. It is possible to treat this deformity with a high tibial osteotomy to realign the knee.[12] Conversely, a valgus knee, which is more common in women, reduces the knee adduction moment and exposes the knee to excessive lateral loading, which can lead to lateral compartmental osteoarthritis.

Learning points

- Muscles have small moment arms, meaning that forces acting on joints *in vivo* during daily activities are large and dominated by muscle forces.
- Operations affecting the size and shape of a joint are likely to impact the muscle moment arms and joint moments, and hence the joint loads — this can have negative effects, for example reducing the femoral offset, or positive effects, for example realigning a varus knee.

Joint loads

When referring to joint loads and moments, it is important to make sure that the intended audience can correctly interpret the results/findings, and to do this a well-defined co-ordinate system is needed. A frequently used co-ordinate system is that recommended by the International Society of Biomechanics, which has defined exact movements and co-ordinate axes for a number of joints.[13]

Experimental measurement of kinematics and contact forces

3D motion analysis techniques, such as optical tracking, are used to record joint kinematics during functional activities such as gait, stair climbing and rowing. Measuring ground reaction forces then also allows the calculation of the joint dynamics (also known as kinetics) from the kinematics. These techniques are widely used for a variety of investigations, ranging from hip joint dislocation[14] to the effects of ACL injury on joint loading.[15]

The current gold standard for measuring the joint reaction forces comes from Bergmann et al.'s instrumented prostheses.[5,6] In this research, small force sensors and wireless transmitters are placed inside joint replacement implants, which allow the joint reaction force to be measure post-operatively during a large variety of activities. All the data collected, videos of the trials and loading profiles are freely available for research use, and include data for the hip, knee, shoulder and spine.[6]

Typical loading profiles during gait

During activities of daily living, the joints have compressive loads acting across them. Figure 4 shows a typical loading profile at the hip joint during gait from one of Bergmann's cases. The loads are much higher during the stance phase of gait, and two peaks can be seen: the first occurs at heel strike and the second at toe off. Generally, the forces at heel strike are higher; however, it is also possible for a patient to exert a greater force on the hip joint at toe off depending on their gait technique. It can also be seen that the maximum load reaches around 2.5 BW, which is due to muscle co-contraction to stabilize the joint at heel strike.

Bergmann's data[6] shows also a similar double peak loading profile at the knee; however, for the knee, the peak at toe off has a tendency to be greater with a similar magnitude to that of the contact force at the hip joint.

Wear

The total wear volume produced by a movement is proportional to the normal load and the sliding distance and inversely proportional to the hardness of the material. Thus, large *in vivo* forces will increase wear rates of cartilage or total joint replacement bearing surfaces. This means that heavier patients are more likely to wear out a joint replacement, as are younger patients because activity levels are also highly correlated with *in vivo* wear rates.[16] Recent trends in hip replacements have seen a move towards harder couples such as ceramic-on-ceramic or metal-on-metal, as these will give lower wear rates for the same loading conditions.

Learning points

- Joint forces in the lower limb frequently exceed bodyweight, and can be as high as ten times bodyweight.
- During the gait cycle, the contact force reaches 2–3 times bodyweight, is much higher during the stance phase of gait, and peaks at heel strike or toe off depending on the joint and patient.
- Wear of articulating surfaces (both natural cartilage and joint replacements) is proportional to the joint loads — excessive loads cause problems.

Contact stresses

The compressive contact forces across joints create contact stresses in the articulating surfaces, and if these stresses are too high then the material can wear, deform or even fracture. The amount of contact stress is proportional to the load divided by the contact area, and thus the risk of articulating surface damage is increased by either abnormally high contact loads or through abnormally small contact areas.

The influence of the knee meniscus on contact stresses

The knee menisci are two crescentic wedge-shaped fibrocartilage structures that are attached to the tibial plateau at the tips of the crescents, see Figure 5a. Their main function is to protect the articular cartilage in load-bearing by increasing the contact area between the femur and tibia, thus protecting the joint against excessive contact stresses, as shown in Figure 5b and c.

The menisci are finely tuned soft tissues. They consist of a porous hydrophilic matrix reinforced with collagen fibres, which allows them to provide a low friction articulating surface similar to articular cartilage. Importantly, however, the meniscus also has collagen fibres aligned in such a way that it can assist greatly in load support: the central portion of the meniscus contains radially orientated fibres so that the meniscus deforms readily under compressive loads, filling the joint space between the condyles (and thus increasing the contact area and decreasing the contact stresses). Conversely, the peripheral fibres are orientated circumferentially, which means that they can resist the extrusion of the meniscus out of the joint space; and by doing so, take the load.[17] If the meniscus is visualized as being part of a circle, the radial extrusion must increase the circumference of the circle as the radius increases — hence the 'hoop' tension around the periphery and the importance of the attachment ligaments.

The fine-tuned structure of the menisci has important implications for meniscal tears, a common painful knee injury. Surgeons traditionally treated meniscal tears with a complete meniscectomy, and whilst this can alleviate pain in the short term, it also increases contact stresses on the adjacent cartilage and hence exposes the joint to premature osteoarthritis. As a result, it is now recommended to only perform a partial meniscectomy, and to repair the meniscus where possible. However, it should be noted that a partial meniscectomy that completely cuts the peripheral circumferential fibres may functionally represent a full meniscectomy, as the tissue is no longer able to share the load.[18]

Edge loading in hip replacements

Typical *in vivo* contact stresses at the hip during gait are around 5 MPa, and high contact stresses occurring during rising from a chair are about 15 MPa.[19] These values are well below the fracture toughness of ceramics, the yield strength of cobalt-chrome (the metal alloy used in metal-on-metal joint replacements), and are too low to cause significant levels of wear in hard-on-hard bearings.

However, under edge loading conditions, the contact area between the femoral head and acetabular shell is greatly reduced, causing contact stresses in the range of 200–1000 MPa.[20] The high contact stresses in an edge-loaded bearing dramatically increase the localized wear rate, leading to bearing material damage. In ceramic-on-ceramic bearings, edge loading causes higher

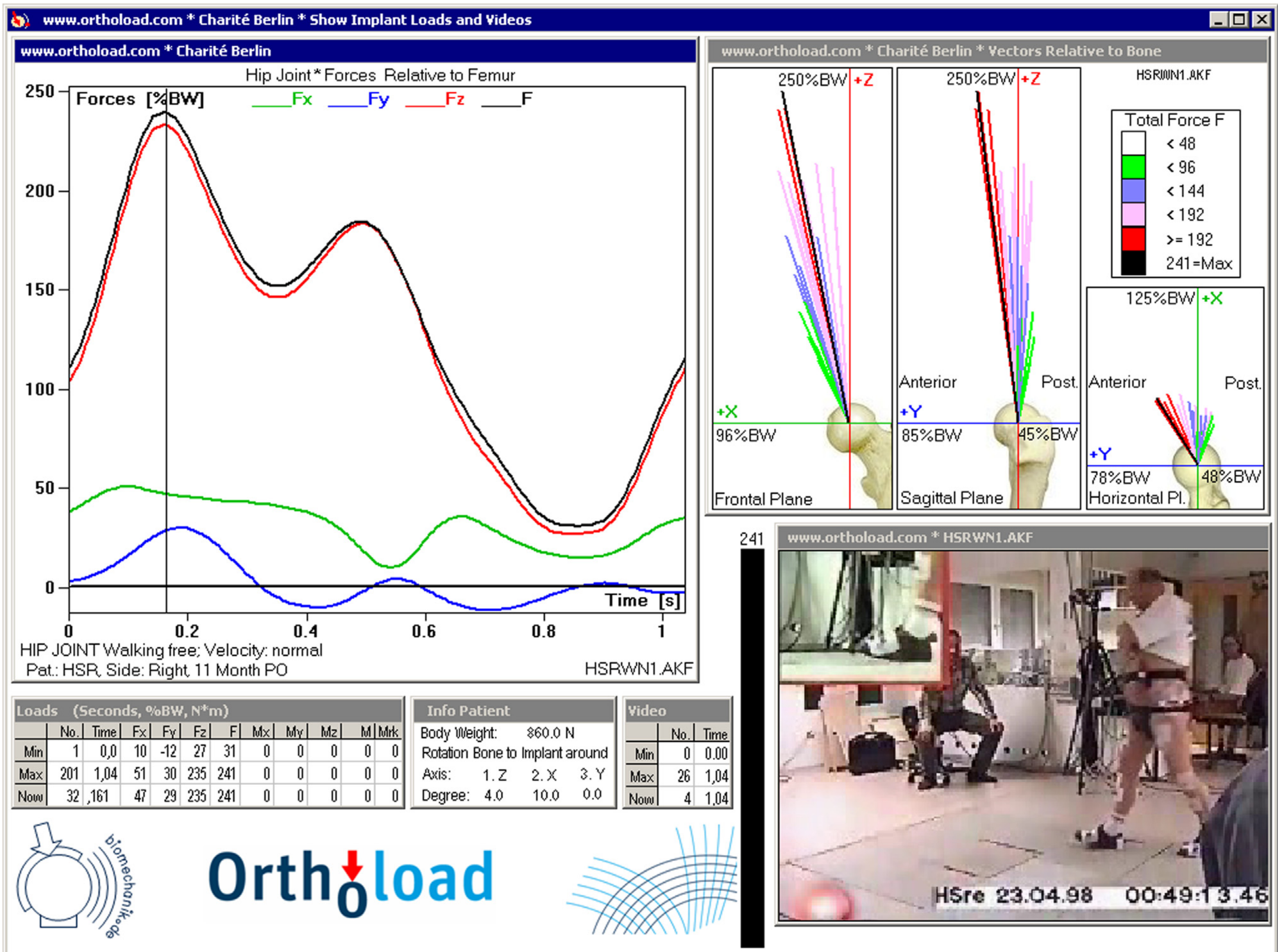

Figure 4 A typical loading profile for the hip during gait (file hsrwn1_fmax.png from database OrthoLoad[6]). Two peaks can be seen; the first occurring at heel strike and the second at toe off.

wear rates but the effects, including squeaking, rarely warrant revision.[21] In large diameter metal-on-metal hip replacements, however, the wear problem is severe: high stresses during edge loading are combined with large sliding distances (due to the large circumference of the bearing), so this leads to much higher wear rates than the prosthesis was designed to see. The large volumes of metal ions released have presented a huge revision burden, as pseudotumours form around the hip joint causing severe pain and poor functional outcomes.[22]

Figure 6 demonstrates the mechanisms of edge loading. It can occur through impingement, followed by a subluxation that exposes the femoral head to the sharp rim of the acetabulum (Figure 6a).[20] Soft tissue laxity (typically caused by a reduced femoral offset) can also lead to edge loading as it allows for

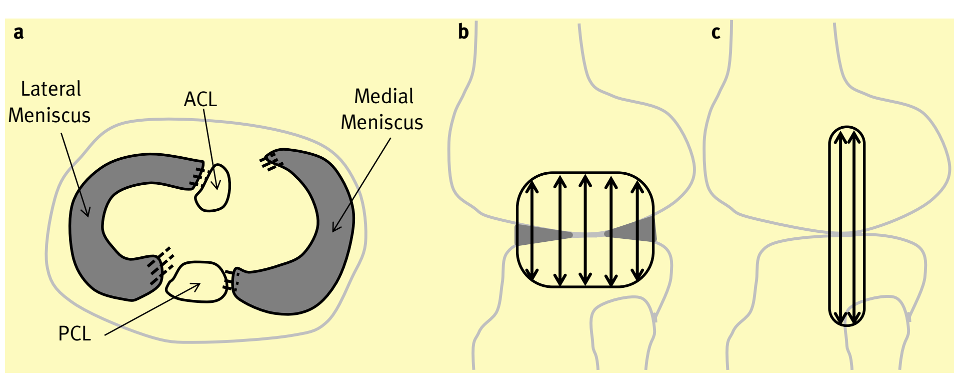

Figure 5 (**a**) The knee meniscus on the tibial plateau. (**b**) A lateral view of a knee with a meniscus under load — it can be seen that the meniscus increases the contact area and thus decreases the contact stresses. (**c**) The same knee under the same load without a meniscus — the contact area is greatly reduced and hence the contact stresses are much larger and more concentrated.

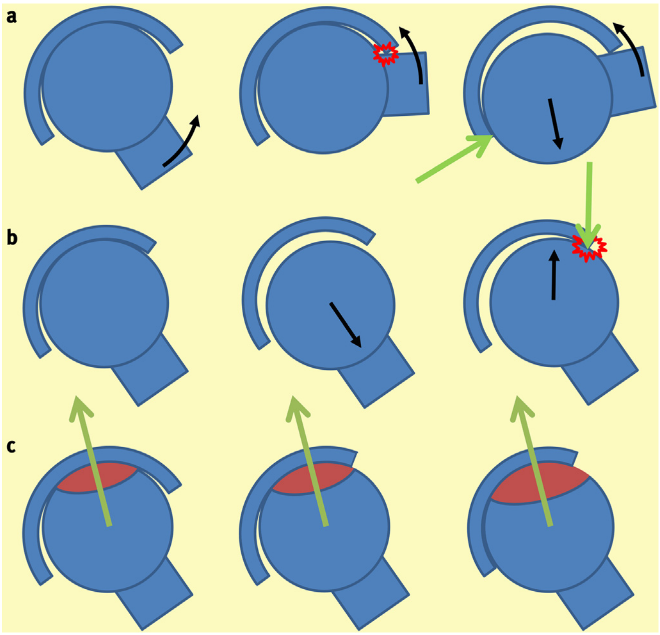

Figure 6 Mechanisms of edge loading in hip replacements. (**a**) Anterior impingement leads to subluxation and posterior edge loading. (**b**) Microseparation during the swing phase of gait leads to edge loading as the bearing relocates on the superior edge during heel strike. (**c**) High cup inclination allows the contact patch to extend past the superolateral acetabular rim during gait. This effect is further amplified by low clearance, which increases the size of the contact patch.

microseparation of the bearing surfaces during the swing phase of gait followed by edge loading as the bearing relocates during heel strike (Figure 6b).[9] Finally, edge loading can occur in the absence of impingement or microseparation if the contact patch caused by the femoral head on the acetabular shell extends past the rim of the acetabulum (meaning that there is insufficient contact area to support the head, increasing the contact stresses). This final mechanism is particularly common in highly inclined bearings, or bearings with reduced subtended angles on the acetabular component, as both these factors bring the superior edge of the cup closer to the path of the contact force vector during activities such as stair climbing and gait (Figure 6c).[3]

This risk of edge loading by these mechanisms can be greatly reduced by the surgeon: careful cup positioning decreases the risk of impingement and insufficient femoral head coverage during gait. Correctly restoring the soft tissue balance with a correct femoral offset can help prevent microseparation during gait.

Articular cartilage

Articular cartilage has little regenerative capacity, yet in many people it can provide a low friction articulating surface capable of transmitting the high *in vivo* contact stresses through the joint for their whole lives. Cartilage is able to do this due to its biphasic properties — it has both a solid and a liquid phase. Its strength is due to collagen fibres, which anchor into subchondral bone, pass through the thickness of the cartilage and are tangential to the articulating surface. This arrangement traps hydrophilic proteoglycan molecules, which attract water by osmosis, creating a tense 'swollen' hydrated structure.

A basic description of the function of articular cartilage is that the solid cartilage matrix is protected against high contact stresses by movement of the fluid between its pores. This movement creates large drag forces, which result in pressure gradients that can support the applied load.[23] In other words, provided the cartilage is well hydrated, the load is supported by the fluid and not the solid matrix, and hence it is protected from damage. Furthermore, articular cartilage provides a low friction surface as solid-on-solid contact between two adjacent articulating surfaces is prevented by a lubricating fluid layer. Unfortunately, surface damage leads to fibrillation, when the proteoglycans escape and the cartilage loses its tensed, hydrated stiffness, and that allows degenerative changes to cascade.

Learning points

- Contact stresses are proportional to loads and inversely proportional to area, and thus are increased by increasing loads or decreasing contact areas.
- The meniscus increases the contact area between the femoral and tibial condyles, and hence reduces the contact stresses — complete or partial meniscectomy at the periphery exposes the cartilage to high contact stresses and early degeneration.
- Edge loading of a hip prosthesis exposes the bearing to extreme contact stresses, but the mechanisms of edge loading can be prevented by careful cup placement and soft tissue balancing.

Joint stability

A stable object is one that returns to its original place when it is displaced. For a joint this would mean the joint returning to or

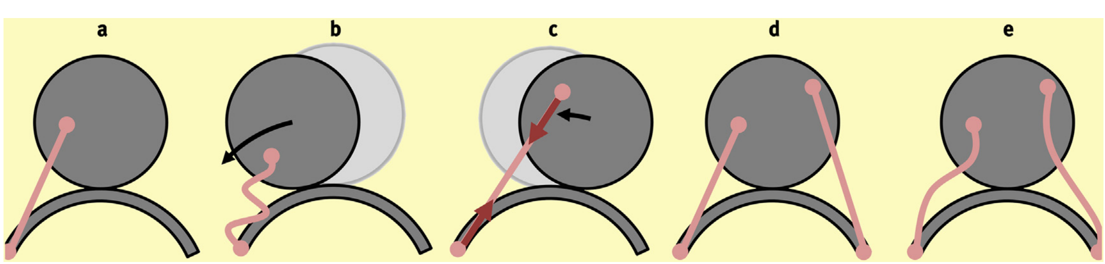

Figure 7 How ligaments provide stability. (**a**) A partially stable configuration. (**b**) When the ball of the joint is rolled to the left then the joint is unstable and the ball is able to continually roll and would permanently dislocate unless relocated by muscle action. (**c**) If the ball is rolled to the right then the ligament provides a passive restoring force that is proportional to the amount of rotation, and this returns the ball to its original position. (**d**) A fully stabilized joint with a limited range of motion. (**e**) A joint protected against dislocation by ligaments, with a larger range of motion than **d**; however, muscular action is needed to enhance joint stability where the ligaments are slack.

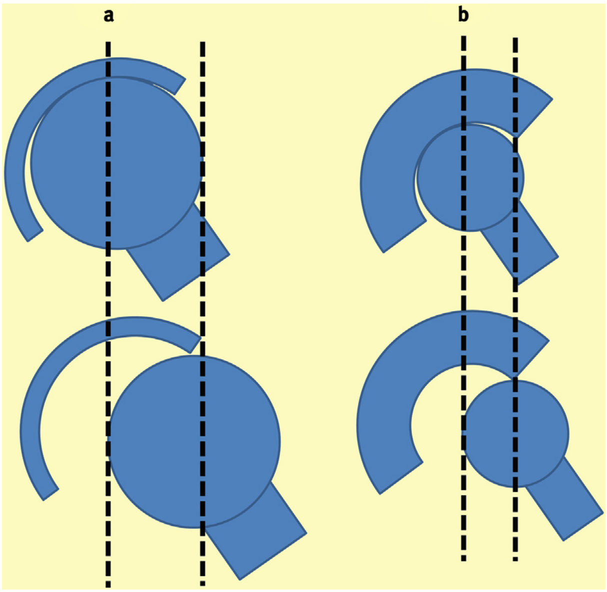

Figure 8 The effect of reducing femoral head size on hip joint stability. (**a**) A normal hip, large diameter total hip replacement or hip resurfacing has a large femoral head and thus there is a large jump distance prior to dislocation. (**b**) Conversely, a small traditional total hip replacement requires only a small amount of subluxation before dislocation can occur.

maintaining its original position upon application of an external load. This stability can be achieved actively or passively; active stability comes from the action of muscles, and passive stability from the joint shape, the ligaments and the fibrocartilage structures. The body makes use of both types of stability, as passive stability is advantageous in terms of energy expenditure and does not require neuromuscular control whereas active stability allows fine control of limb position over a large range of motion. The amount of restoring force exerted by a ligament, and hence its contribution to stability, is proportional to its increase in length during the movement. Figure 7a–e shows how ligaments can be used to provide stability to a joint.

Passive stability at the hip — femoral head size and capsular repair

The hip joint is naturally stable because the acetabulum and femoral head have congruent shape, and thus the femur readily relocates back in the acetabulum when displaced. In total hip arthroplasty, the femoral head size is often reduced, which exposes the joint to an increased risk of dislocation,[24,25] (Figure 8) and as a result implant manufacturers have worked towards increasing the femoral head size. The main hurdle faced by a larger head size is that larger heads increase wear rates because of the increased sliding distance for an equivalent change in rotation, caused by the larger circumference of the bearing.

Passive stability at the knee — joint shape

In the knee, the medial condyles are marginally congruent; the femoral condyle is convex and the tibial condyle concave, and hence this provides some passive stability. The lateral compartment, however, is naturally unstable, with both the femoral and tibial condyles having a convex shape. This fine-tuned joint shape arises from conflicting requirements within the knee. The medial condyle carries a greater proportion of the load due to the adduction moment at the knee, so the congruent joint shape aids load transfer by increasing the contact area and decreasing the contact stresses. By contrast, the morphology of the lateral compartment enables a greater range of knee motion, as in deep flexion the lateral condyle's shape permits small amounts of tibial internal rotation, which moves the lateral tibial plateau anteriorly under the femur and thus increases the impingement free range of motion. Mimicking this medial rotation in deep flexion is a goal for modern total knee replacements.[26]

Passive stability at the knee — the ACL

The anterior cruciate ligament (ACL), which attaches anteriorly on the tibia and posteriorly on the femur, and which passes between the condyles, prevents the tibia sliding anteriorly under the femur (or the femur posteriorly over the tibia). The ACL is the 'primary restraint' to tibial anterior translation, resisting approximately 90% of the displacing force. Other structures such as the medial collateral ligament (MCL) make small contributions, and are therefore referred to as secondary restraints to anterior translation. The ACL is aligned such that it is stretched directly when the tibia moves anteriorly, and so elastic tension rises rapidly (Figure 9a). Conversely, the MCL is orientated perpendicular to this subluxation, so it is not stretched very much.[27]

The ligaments around the knee must stabilize it against all the loads encountered. When the load causes tibial abduction (a valgus moment), the tibia pivots about the lateral femoral condyle. Then, the MCL, which has the largest moment arm,

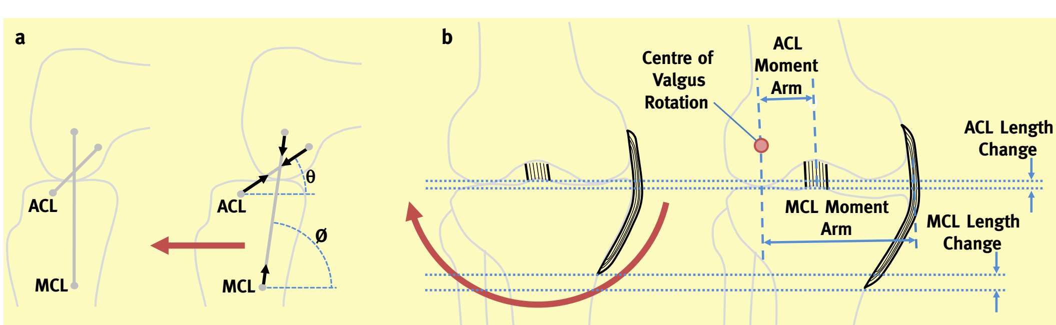

Figure 9 Primary and secondary stability at the knee. (**a**) During anterior drawer, the ACL provides the primary restraint; the ACL is aligned more with the direction of the translation than the MCL and so undergoes larger amounts of extension (resulting in greater tension in the ligament), and it has a larger component of force in the anterior direction (θ is smaller than $\varnothing$). (**b**) Conversely, under a tibial abduction moment, the MCL acts as the primary restraint due to its larger moment arm and larger extension.

becomes the primary restraint. The ACL is closer to the lateral condyle, so it is not stretched much and has a small moment arm, so it is only a secondary restraint to tibial abduction (Figure 9b).

During knee flexion, the ACL greatly affects the knee joint kinematics. Initially, as the knee flexes the ligament slackens, allowing the femur to roll posteriorly over the tibial plateau. This roll back shifts the axis of rotation posteriorly, which increases the quadriceps moment arm and hence decreases the muscle force required to extend the knee again. The ACL tightens with further knee flexion, and the kinematics change from a roll back to a continued rolling plus sliding motion, and so the ACL prevents dislocation whilst allowing deep knee flexion.[28] The posterior cruciate ligament (PCL) acts in a reciprocal manner during knee extension.

The ACL also has an additional function in knee extension, as it tightens as part of the 'screw home' mechanism of the knee. This tightening of the knee ligaments stiffens the joint in extension, and hence less muscle energy expenditure is needed to stabilize the knee whilst standing upright or during heel strike in the gait cycle.

Learning points

- The soft tissues around any joint are carefully balanced and protect it against instability. Joint instability can lead to abnormal loading and degenerative conditions such as osteoarthritis.
- The hip joint is naturally stable, but reducing the femoral head size and damaging the posterior capsule during hip arthroplasty increases the risk of instability and means that dislocation can occur.
- The knee joint is inherently less stable than the hip, so relies greatly on ligaments to maintain stability. There are primary and secondary restraints to each movement, and their integrity is examined by routine clinical tests such as anterior-posterior drawer tests.

Conclusion

This article has shown how the equilibrium of human joints can be analyzed by considering the balance between the moments caused by external forces, the bodyweight and the muscles. The muscles have small moment arms, so the force they exert must be large. This leads to lower limb joint forces several times bodyweight when walking. The articular congruence of the joint surfaces spreads the joint force across a large contact area, which reduces the contact stresses. Trauma that disrupts this congruence, such as meniscal damage, leads to elevated articular cartilage stresses and thus to degenerative changes. ◆

REFERENCES

1 Baliunas AJ, Karrar A, Hurwitz DE, et al. Increased knee joint loads during walking are present in subjects with knee osteoarthritis. *Osteoarthritis Cartil* 2002; **10:** 573–9.

2 Lohmander LS, Englund PM, Dahl L, Roos E. The long-term consequence of anterior cruciate ligament and meniscus injuries: osteoarthritis. *Am J Sports Med* 2007; **35:** 1756–69.

3 Kwon Y-M, Mellon SJ, Monk P, Murray DW, Gill HS. In vivo evaluation of edge-loading in metal-on-metal hip resurfacing patients with pseudotumours. *Bone Joint Res* 2012; **1:** 42–9.

4 Charles MN, Bourne RB, Davey JR, Greenwald AS, Morrey BF. Soft-tissue balancing of the hip - the role of femoral offset restoration. *J Bone Joint Surg Am* 2004; **86A:** 1078–88.

5 Bergmann G, Deuretzbacher G, Heller M, et al. Hip contact forces and gait patterns from routine activities. *J Biomech* 2001; **34:** 859–71.

6 Bergmann G, ed. Charite – Universitaetsmedizin Berlin (2008). OrthoLoad. Retrieved (3 Jan 2013) from: http://www.OrthoLoad.com

7 Sakalkale DP, Eng K, Sharkey PF, Hozack WJ, Rothman RH. Effect of femoral component offset on polyethylene wear in total hip arthroplasty. *Clin Orthop Relat Res* 2001; 125–34.

8 McGrory B, Morrey B, Cahalan T, An K, Cabanela M. Effect of femoral offset on range of motion and abductor muscle strength after total hip arthroplasty. *J Bone Joint Surg Br* 1995; **77-B:** 865–9.

9 Lombardi AV, Mallory TH, Dennis DA, Komistek RD, Fada RA, Northcut EJ. An in vivo determination of total hip arthroplasty pistoning during activity. *J Arthroplasty* 2000; **15:** 702–9.

10 Zhao D, Banks S, Mitchell K, D'Lima D, Colwell C, Fregly B. Correlation between the knee adduction torque and medial contact force for a variety of gait patterns. *J Orthop Res* 2007; **25:** 789–97.

11 Shelburne K, Torry M, Pandy M. Contributions of muscles, ligaments, and the ground-reaction force to tibiofemoral joint loading during normal gait. *J Orthop Res* 2006; **24:** 1983–90.

12 Sprenger T, Doerzbacher J. Tibial osteotomy for the treatment of varus gonarthrosis. Survival and failure analysis to twenty-two years. *J Bone Joint Surg Am* 2003; **85-A:** 469–74.

13 Wu G, Siegler S, Allard P, et al. ISB recommendation on definitions of joint coordinate system of various joints for the reporting of human joint motion – part I: ankle, hip, and spine. International Society of Biomechanics. *J Biomech* 2002; **35:** 543–8.

14 Nadzadi ME, Pedersen DR, Yack HJ, Callaghan JJ, Brown TD. Kinematics, kinetics, and finite element analysis of commonplace maneuvers at risk for total hip dislocation. *J Biomech* 2003; **36:** 577–91.

15 Hart J, Ko J-W, Konold T, Pietrosimione B, Pietrosimone B. Sagittal plane knee joint moments following anterior cruciate ligament injury and reconstruction: a systematic review. *Clin Biomech* 2010; **25:** 277–83.

16 Schmalzried TP, dela Rosa M, Shepherd EF, et al. The John Charnley Award. Wear is a function of use, not time. *Clin Orthop Relat Res* 2000; **381:** 36–46.

17 Masouros SD, McDermott ID, Amis AA, Bull AMJ. Biomechanics of the meniscus-meniscal ligament construct of the knee. *Knee Surg Sports Traumatol Arthrosc* 2008; **16:** 1121–32.

18 McDermott ID, Amis AA. The consequences of meniscectomy. *J Bone Joint Surg Br* 2006; **88:** 1549–56.

19 Hodge WA, Carlson KL, Fijan RS, et al. Contact pressures from an instrumented hip endoprosthesis. *J Bone Joint Surg Am* 1989; **71:** 1378–86.

20 Elkins JM, O'Brien MK, Stroud NJ, Pedersen DR, Callaghan JJ, Brown TD. Hard-on-hard total hip impingement causes extreme contact stress concentrations. *Clin Orthop Relat Res* 2011; **469:** 454–63.

21 Esposito CI, Walter WL, Roques A, et al. Wear in alumina-on-alumina ceramic total hip replacements: a retrieval analysis of edge loading. *J Bone Joint Surg Br* 2012; **94-B:** 901–7.

22 Langton DJ, Joyce TJ, Jameson SS, et al. Adverse reaction to metal debris following hip resurfacing: the influence of component type, orientation and volumetric wear. *J Bone Joint Surg Br* 2011; **93:** 164–71.

23 Katta J, Jin Z, Ingham E, Fisher J. Biotribology of articular cartilage — a review of the recent advances. *Med Eng Phys* 2008; **30:** 1349—63.

24 Bartz RL, Noble PC, Kadakia NR, Tullos HS. The effect of femoral component head size on posterior dislocation of the artificial hip joint. *J Bone Joint Surg Am* 2000; **82A:** 1300—7.

25 Berry D, von Knoch M, Schleck C, Harmsen W. Effect of femoral head diameter and operative approach on risk of dislocation after primary total hip arthroplasty. *J Bone Joint Surg Am* 2005; **87:** 2456—63.

26 Fitz W, Sodha S, Reichmann W, Minas T. Does a modified gap-balancing technique result in medial-pivot knee kinematics in cruciate-retaining total knee arthroplasty? A pilot study. *Clin Orthop Relat Res* 2012; **470:** 91—8.

27 Butler DL, Noyes FR, Grood ES. Ligamentous restraints to anterior-posterior drawer in the human knee. A biomechanical study. *J Bone Joint Surg Am* 1980; **62:** 259—70.

28 Biomechanics of the knee. In: Standring S, ed. Gray's anatomy. 40th edn. London: Churchill Livingstone Elsevier, 2009; 1403—1407.

