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Basic Biomechanics of the Hip

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(1)Abstract

The basic mechanical principles which govern how the hip joint maintains equilibrium and balance during standing and performing activities is explained along with the consequences when this balanced system is compromised. A description of the movements and forces acting around the hip joint that are expected during activities of daily living is offered and also how these movements are affected following total hip replacement, with particular reference to femoral offset and leg length inequality.

Keywords

Hip Biomechanics; Total Hip Replacements; Activities of Daily Living; Femoral Offset; Leg Length Inequality

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(1)Introduction

An understanding of the mechanics of the hip joint is important background knowledge for a number of disciplines, whether these are for the diagnosis and treatment by a clinician, or for the surgeon who is performing hip surgery. It is important to understand how the mechanics of the hip change when a person is static, compared to when dynamic, what anatomical structures interact and how these enable movement and maintain stability within these mechanical principles. It is also important to understand what the normal function of the hip is during activities of daily living and how these are changed when hip surgery has been performed.

(1)Hip Anatomy

The hip joint is surrounded by a mass of musculature that produces desired movements at both the hip and the knee, and prevents unwanted movements from the inertial forces caused by the large moving masses. The resulting joint reaction force at the hip can thus exceed many times our body weight demonstrating the importance of soft tissue support [1,2].

Muscles, however, are not the only important soft tissue structures which influence the integrity of the hip joint. The hip joint has a strong joint capsule and is surrounded by a complex ligamentous structure. The joint capsule has a protective role to restrain the movement of the femur articulating around the acetabulum and to prevent dislocation. The extracapsular ligaments comprise the iliofemoral (IF) pubofemoral (PF) and ischiofemoral (ISF) ligaments. The iliofemoral ligament is a Y-shaped ligament which extends anteriorly from the ilium, attaching to the intertrochanteric line of the femur; the IF ligament prevents hyperextension of the hip. The PF ligament attaches to the obturator crest and superior ramus of the pubis and blends with the articular capsule; the PF ligament prevents excessive abduction and extension of the hip. The ISF ligament is located on the posterior aspect of the hip, originating from the ischium and inserting on the intertrochanteric line of the femur; the ISF ligament resists hip hyperextension and excessive internal rotation.

Ligaments are passive structures and act more like a resistance band, thus it is very difficult to quantitatively measure when and how they work. In contrast, muscular activity can be measured through methods such as electromyography (EMG). Because of this, in comparison to studies orientated around muscle function, there is far less research around the mechanical influence of the ligaments. Researchers have used cadaveric hips to measure the contribution of each ligament using range of motion (ROM) testing, by measuring the reduction in measured torque when the ligaments had been removed [3]. In this manner the IF and ISF ligaments were found to have an essential role in restraining rotational hip movement. Thus to maintain correct hip mechanics following surgery it is important that the function/tension of the individual ligaments is considered. To truly understand the role that soft tissues play it is important to look at how the hip joint functions as a supportive and mobile structure.

(1)Mechanics of the Hip Joint

The human body is a well-engineered structure where bone and soft tissues interact in both static and dynamic situations to maintain balance and generate movement. Statics is a branch of mechanics which models and analyses load on a physical system, where structures are motionless or moving at a constant velocity. Such models would include the hip joint when standing still.

During static standing the combined forces acting on any component, measured in Newtons (N), must be zero in all in all axes (Figure 1), thus for translational static equilibrium:

$$\sum F_{x,y,z} = 0 \qquad \text{(Equation 1)}$$

Where F= force and x, y, z are the axis of rotation.

Torque, measured in units of Newton.metres (Nm), at the hip joint is also experienced and is the consequence of a load acting at a distance. For rotational static equilibrium the sum of the moments needs also to be zero [4]:

$$\sum M_{x,y,z} = 0 \qquad \text{(Equation 2)}$$

Where M= moments

During walking the leading leg leaves the ground to step forwards, thus temporarily the body is standing on one leg. The force from our body weight (BW) at this time acts downwards pulling the body to lean over, however, this is balanced by the action of the abductors. Thus the hip behaves much like a lever (Figure 1), with a load/effort acting either side of a fulcrum (femoral head).

During standing, however, BW is supported by both hips, therefore, if the body was perfectly balanced the abductor muscles would not be required and there would be an equal force of ½ BW on each hip. As it is unlikely that the body is ever perfectly balanced the joint reaction force during standing likely varies from ½ BW to 3BW, for the perfectly balanced case and single leg stance case respectively. The abductor muscles are thus very important in balance and pelvic stability, their role becoming more important as motion becomes more dynamic. It is worth noting that although 2D static analysis provides a realistic estimation of forces and moments a number of assumptions are necessary, these are listed in Table 1.

During gait there are two distinctive phases the stance phase, when the foot is contact with the floor and the swing phase when the leg is returning. The role of the abductors is that of balance on the loaded side and managing limb motion on the unloaded side as the leg is brought forward. From the simplistic representation in Figure 1 the abductor muscle force (ABD) on the stance side is equal to body weight multiplied by the ratio of the moment arms of BW (b) and the abductors (a) measured from the hip joint centre (Equation 3).

$$ABD = BW \times b/a \qquad \text{(Equation 3)}$$

Where ABD= abductor muscle force; BW= Bodyweight; b = body weight moment arm; a = abductor moment arm.

The application of force by the abductor muscles means that the hip is never totally unloaded even when no BW is being applied, as movement of the mass of the leg during the swing phase requires muscles to control this motion. The moment (force of the abductor muscle x moment arm of the abductor muscle) applied by the abductors relative to the hip centre during gait is shown in Figure 2. It is clear that during stance the magnitude of the moment applied by the abductors is far greater than during swing. The understanding of these mechanics is important when understanding pathologies which might change the length of the muscle moment arms, as if moment arms shorten to achieve the same moment the muscle force must increase. Thus if the mechanics changes considerably then the patient will have to adopt coping strategies to maintain balance and equilibrium.

If a hip becomes painful due to arthritis then the pain can be alleviated by reducing the joint reaction force. From Equation 3, this can be achieved by a reduced BW moment arm (b) if the patient leans towards the painful hip so that the abductor muscle can apply a reduced force to achieve stability. The same thing happens if we stand on one leg as we tend to try to get our BW centred above the hip so it requires the least amount of force from the stabilising musculature. Alternatively to alleviate hip pain is to use a walking stick on the opposite side of the painful hip (Figure 1); this reduces the hip abductor force and thus can reduce the joint contact force and pain in the affected limb. The previously used equation 3 now becomes:

$$ABD = (BW \times b/a) - (WS \times c/a) \qquad \text{(Equation 4)}$$

Where ABD= abductor muscle force; BW= Bodyweight; b = body weight moment arm; a = abductor moment arm; WS= walking stick force; c= walking stick moment arm.

Clinically, abductor weakness often leads to a characteristic drop in the pelvis during the stance phase of walking, to the non-weight bearing side, referred to as a Trendelenburg gait. A similar tilt of the pelvis to the opposite side during single leg stance is referred to as a positive Trendelenburg sign. This should not be confused with the Trendelenburg test (or Brodie-Trendelenburg test), which is a test of leg vein competency, although the terms sign and test are often used interchangeably in textbooks.

When weakness occurs on one side, compensating movement of the body may change the direction of load, transferring forces further down the kinetic chain to other joints. For example, to compensate for pelvic drop during Trendelenberg gait, the knee of the contralateral limb may go into a valgus/rotated position. This is recognised as a risk for knee injury, as well as arthritis, due to excessive shear forces acting on the knee joint [5 6]. Therefore maintaining the balance of the pelvis is an important consideration for the clinician when protecting other joints as well as the affected joint.

(2)The Hip during Activities of Daily Living

Motion capture allows for a comprehensive analysis of the movements performed during gait that may subsequently be used to calculate muscle/joint forces. Motion capture is frequently performed in a lab which generally would contain a number of infrared cameras to capture movement and force plates to measure ground reaction forces. During gait analysis one single gait cycle is typically normalised to 100%, with one cycle beginning with a heel strike and ending the next time the same heel makes contact with the ground, with a toe off event at 60% of the gait cycle. The cycle can then be subsequently broken down into subsections and events such as stance and swing, heel strike and toe off, this can be seen in Figure 3.

When discussing hip movements we refer to the femoral movement in relation to the pelvis around the hip joint centre. The hip allows for a large ROM in all 3 planes allowing for 120° flexion/10°extension, 70° abduction/adduction and 50° rotation, these movements are depicted in Figure 4 [7]. These ranges are the maximum angles that the hip can safely achieve, however these ranges are rarely reached during activities of daily living, hence under normal activities muscles are generally responsible for providing all of the rotational stability.

Normal kinematics of the hip during level gait (Figure 5) reveals a large ROM in the frontal and transverse planes, in comparison to other joints[8]. During the gait cycle (Figure 4) the hip is flexed at the initial heel contact of the stance phase before hip joint begins to extend until the end of the stance phase where flexion begins. This is coupled with hip abduction during mid-stance when the hip begins to abduct until the end of stance phase prior to adduction until the end of the gait cycle.

In the majority of past studies, gait has been used as the primary activity to analyse kinematics of the hip joint. However, studies involving total hip replacement have highlighted functional demand as an important outcome measure for patient satisfaction. Thus recently there has been growing interest in activities of daily living (ADLs) to obtain a true representation of how the hip moves on a day to day basis. The ADLs which are analysed are often more demanding than gait by requiring increases in ranges of motion and/or joint moments. The typical activities of daily living which are often analysed are an increased walking speed, a sit to stand task, and ascending and descending stairs.

The sit-to-stand (STS) task is performed ~60 times a day by healthy adults and as an activity of daily living is unusual in its movement (Table 2) [9]. Most activities which are performed during daily living are performed in a unilateral pattern whereas the STS task is performed identically bilaterally. Furthermore the high degree of flexion at the hip at the start of a STS task makes the movement challenging both for maintaining balance and for producing the force needed to complete the movement. Following joint replacement, completion of this task represents the mechanical efficiency of the quadriceps muscle and how well the associated moment arms have been reconstructed. Stair ascending creates a greater demand on the muscles compared to descending which is much more about control of the movement than force production. These increases in demand can be seen when considering the increased flexion angle achieved during the sit-to-stand and stair ascent tasks compared to normal walking (34° and 66° respectively).

The different demands placed on the hip joint during the ADLs are apparent in Table 2. The surprising differences between the moments are the relatively similar values between the STS and stair ascent/descent tasks. It would be expected that due to the impact nature of stair descent hip moments would be high. Moments provide information that helps us understand the nature of the forces acting around the joint. However it is also important to understand what is happening at the surface contact interface between the acetabulum and the femoral head. Figure 6 shows a typical hip joint reaction force during the gait cycle with the initial peak occurring just after the first heel strike followed by a second peak just before the toe off.

Measuring joint contact forces is difficult by the nature of the task. There have been a few studies, however, that have implanted instrumented prostheses to measure hip contact forces during different tasks [2]. The results of these studies are summarised in Figure 7 demonstrating how joint contact forces change during different activities. The Bergmann study found that the largest contact force was measured during the stair descending trials, reaching 260% bodyweight and fast walking was the second highest (Figure 7). The lowest joint contact force was produced during the sit to stand activity.

The alternative to instrumented prostheses are computational joint contact force models that utilise multi-body inverse dynamics. The results of these studies are difficult to validate but are generally comparable to that of Bergmann et al [10]. The advantage of software simulation is that they are less invasive and allow for a fairly rapid acquisition of data, thus facilitating large datasets for a better statistical representation of variation within a given population.

(2)Effect of Total Hip Replacement Surgery

The aim of total hip replacement (THR) surgery is to reduce pain and restore normal function. Despite hip surgery being common place the procedure is still a significant event for the patient due to its invasive nature. It addition there is potential for variation due to the different surgical approaches utilised and the fact that implant positioning can greatly influence the resulting biomechanics of the hip.

Gait analysis following THR has shown that a number of walking parameters are affected compared to healthy control patients including walking velocity, ROM and joint moments (Figure 8). In a recent review, THR patients were compared to that of healthy control patients [11]. It was found that in almost all studies there was reduced ROM in the hip following THR, compared to control subjects including a reduced hip abduction moment and reduced hip abduction angle [12]. It was stated that this would shift the bodyweight over the operated limb, thus reducing the muscle moment required to stabilise the pelvis. This shift might be necessary to compensate for abductor weakness, which could be either due to the surgical procedure or as a residual weakness caused by pain avoidance prior to the hip surgery.

To investigate differences further it is important to consider more demanding activities. Lamontagne et al. measured the differences between THR patients and healthy controls during sitting and standing tasks [14]. They found that the differences occurred at the more demanding part of ADL such as the beginning of the standing phase and the end of the sitting phase of STS. The main differences were a reduction in the hip extension moment.

Similarly, when negotiating stairs, differences have been observed. However in a review by Kolk et al. it was suggested these differences are not as apparent as those observed during level walking[11]. It has been suggested that this is due to the reduced hip joint moments required for stair negotiation when compared to level walking [13]. However this does not provide a full insight into hip mechanics, as the reduced hip moment occurred in combination with high joint contact forces acting at the hip during stair negotiations compared to other activities. Therefore the hip moments may not be a true representative of the joint work load and muscle activity. These results are all based on patients who have had a successful hip operation; in reality this is not always the case and patients can, although being mostly pain free compared to pre-surgery, have poor outcomes such as a compromised gait, limping or leg length inequalities.

(3)Placement of the Implants

Restoration of the centre of rotation of the hip joint is an important goal of THR to ensure normal gait and function. Correct use and selection of implants can restore the biomechanics of the hip with appropriate femoral offset and leg length. Modularity of the prosthetic designs offers many options for the surgeon to optimize leg length and femoral offset to match the contralateral hip side [14].

Several methods have been described to measure offset. Femoral offset is generally measured on a standard anterior/posterior pelvis radiograph and is defined as the perpendicular distance from the femoral head centre of rotation to the long axis of the femur. This measurement should be accurately performed and varies according to the hip rotation (Figure 9). The acetabular offset is the horizontal distance from the centre of the femoral head to the midline of pubic symphysis. Some authors define acetabular offset as the distance from the centre of rotation of the femoral head to the inner wall of the quadrilateral plate also called true floor of the acetabulum. The global offset is the addition of the femoral and acetabular offsets, [15] or by measuring CDE in Figure 9. Failure to accurately reconstruct the femoral and global offset may result in impingement, hip instability, polyethylene wear and trochanteric pain.

Higher offset stems create a larger abductor moment arm and this may decrease the hip joint reaction force through a corresponding reduction in the abductor force. This may also be combined with the advantage of a decreased risk of impingement and the disadvantage of increased soft tissue tension, with the potential for trochanteric pain. Intentional increase of femoral offset (Figure 10) is used sometimes to augment hip stability with the disadvantage of the potential for trochanteric bursitis and gluteal pain in 15% of patients at a follow up of 2 to 5 years postoperatively [16]. In contrast decreasing the femoral offset may lead to increased hip joint reaction forces, instability, abductor weakness and gluteus medius lurch, Table 3 summarises the techniques which can be used to increase femoral offset and reduce soft tissue tension therefore decreasing joint reaction force.

Correction of limb length inequality (LLI) without compromising hip stability remains one of the intraoperative challenges in THR. The incidence is difficult to ascertain but evidence suggests that some lengthening occurs in as many as 30% of patients following THR, due most commonly to mal-positioning of the femoral component, and less commonly the acetabular component [17]. When this difference exceeds 20mm, it is more likely to become clinically significant [18,19]. Symptomatic LLI accounts for 8.7% of THR related claims made against the UK Health Service Litigation Authority[20].

Historically there have been two popular preoperative methods for assessment of leg length, the Woolson, and Williamson techniques [21,22]. These techniques utilized anatomical landmarks on the acetabular (inter-teardrop line, inter-ischial line) and femoral (lesser trochanter) sides from a pelvic AP radiograph to assess changes in leg length. However, these methods do not differentiate whether the cause of the inequality was on the acetabular or femoral side. More recently McWilliams et al. modified this by adding a landmark common to both sides, the hip joint centre [28]. This allowed refinement of the assessment of leg length by providing individual leg length measures for the acetabular (C-cup side) and femoral (S-stem side) sides along with an overall (O) leg length as demonstrated in Figure 11.

The majority of leg length inequality patients are asymptomatic. However it can result in groin pain, back pain, abnormal gait or sciatic nerve palsy, therefore affecting patient-related outcomes (PROMs) after THR. [23-26]

A recent review has identified that smaller females are more likely to be susceptible to leg length changes.[20] The reason for this is that a given magnitude of inequality will cause a corresponding change in angle of the pelvis with respect to the sagittal plane. The magnitude of this pelvic angle is inversely proportional to the width of the pelvis, making smaller people more likely to be symptomatic. Changes in pelvic angle can lead to back pain in the long term. In the short term localised changes in hip biomechanics reduce the efficiency of muscles around the hip; for patients with symptomatic leg length inequality this results in a reduced range of motion (Figure 12)[27] . Figure 12 shows the reduction in hip reaction force caused by a characteristic reduced range of motion following leg lengthening, with a notable reduction in hip extension during gait. It should be noted that these data are for symptomatic patients and that many patients (generally taller) cope with changes in leg length better than others [10,28].

(1)Conclusion

The hip is a complex ball and socket joint that articulates with the aid of muscles and bony framework governed by the laws of mechanics much like any other structure. Changes to the joint position, muscles, or framework will lead to an imbalance that must be compensated for by the body, clinically most often leading to a reduction in the range of motion or transfer of load to another limb. Thus the restoration of the hip back to its normal state is important to prevent potentially symptomatic changes in gait.

(1)References

1. Paul J. Paper 8: Forces Transmitted by Joints in the Human Body. Paper presented at: Proceedings of the Institution of Mechanical Engineers, Conference Proceedings1966.
2. Bergmann G, Deuretzbacher G, Heller M, et al. Hip contact forces and gait patterns from routine activities. Journal of Biomechanics. 2001;34(7):859-871.
3. van Arkel RJ, Amis AA, Cobb JP, Jeffers JR. The capsular ligaments provide more hip rotational restraint than the acetabular labrum and the ligamentum teres : an experimental study. The bone & joint journal. 2015;97-b(4):484-491.
4. Beer FP, Johnston E, DeWolf J. Mechanics of Materials, McGraw-Hill. New York. 1992.
5. Powers CM. The Influence of Abnormal Hip Mechanics on Knee Injury: A Biomechanical Perspective. Journal of Orthopaedic & Sports Physical Therapy. 2010;40(2):42-51.

6. Trepczynski A, Kutzner I, Bergmann G, Taylor WR, Heller MO. Modulation of the relationship between external knee adduction moments and medial joint contact forces across subjects and activities. Arthritis & rheumatology (Hoboken, N.J.). 2014;66(5):1218-1227.
7. Bowman Jr KF, Fox J, Sekiya JK. A Clinically Relevant Review of Hip Biomechanics. Arthroscopy: The Journal of Arthroscopic & Related Surgery. 2010;26(8):1118-1129.
8. Bovi G, Rabuffetti M, Mazzoleni P, Ferrarin M. A multiple-task gait analysis approach: kinematic, kinetic and EMG reference data for healthy young and adult subjects. Gait Posture. 2011;33(1):6-13.
9. Dall PM, Kerr A. Frequency of the sit to stand task: An observational study of free-living adults. Applied ergonomics. 2010;41(1):58-61.
10. Li J, Redmond AC, Jin Z, Fisher J, Stone MH, Stewart TD. Hip contact forces in asymptomatic total hip replacement patients differ from normal healthy individuals: Implications for preclinical testing. Clinical biomechanics (Bristol, Avon). 2014;29(7):747-751.
11. Kolk S, Minten MJ, van Bon GE, et al. Gait and gait-related activities of daily living after total hip arthroplasty: A systematic review. Clinical Biomechanics. 2014;29:705-718
12. Beaulieu ML, Lamontagne M, Beaulé PE. Lower limb biomechanics during gait do not return to normal following total hip arthroplasty. Gait & Posture. 2010;32(2):269-273.
13. Nadeau S, McFadyen BJ, Malouin F. Frontal and sagittal plane analyses of the stair climbing task in healthy adults aged over 40 years: what are the challenges compared to level walking? Clinical Biomechanics. 2003;18(10):950-959.
14. Bourne RB, Rorabeck CH. Soft tissue balancing: The hip. The Journal of arthroplasty. 2002;17(4, Supplement 1):17-22.
15. Dastane M, Dorr LD, Tarwala R, Wan Z. Hip offset in total hip arthroplasty: quantitative measurement with navigation. Clin Orthop Relat Res. 2011;469(2):429-436.
16. Incavo SJ, Havener T, Benson E, McGrory BJ, Coughlin KM, Beynnon BD. Efforts to improve cementless femoral stems in THR: 2- to 5-year follow-up of a high-offset femoral stem with distal stem modification (Secur-Fit Plus). The Journal of arthroplasty. 2004;19(1):61-67.
17. Wylde V, Whitehouse SL, Taylor AH, Pattison GT, Bannister GC, Blom AW. Prevalence and functional impact of patient-perceived leg length discrepancy after hip replacement. International Orthopaedics. 2008;33(4):905-909.
18. Knutson GA. Anatomic and functional leg-length inequality: a review and recommendation for clinical decision-making. Part I, anatomic leg-length inequality: prevalence, magnitude, effects and clinical significance. Chiropractic & Manual Therapies. 2005;13(1):11.
19. Guichet J-M, Spivak JM, Trouilloud P, Grammont PM. Lower Limb-Length Discrepancy: An Epidemiologic Study. Clinical Orthopaedics and Related Research. 1991;272:235-241.
20. McWilliams AB, Douglas SL, Redmond AC, et al. Litigation after hip and knee replacement in the National Health Service. Bone & Joint Journal. 2013;95-B(1):122-126.
21. Meermans G, Malik A, Witt J, Haddad F. Preoperative Radiographic Assessment of Limb-length Discrepancy in Total Hip Arthroplasty. Clin Orthop Relat Res. 2010;469(6):1677-1682.
22. Woolson ST, Hartford JM, Sawyer A. Results of a method of leg-length equalization for patients undergoing primary total hip replacement. The Journal of arthroplasty. 1999;14(2):159-164.
23. Mihalko WM, Phillips MJ, Krackow KA. Acute Sciatic and Femoral Neuritis Following Total Hip Arthroplasty. The Journal of Bone & Joint Surgery. 2001;83(4):589-589.
24. Giles LG, Taylor JR. Low-back pain associated with leg length inequality. Spine (Phila Pa 1976). 1981;6(5):510-521.
25. Gurney B, Mermier C, Robergs R, Gibson A, Rivero D. Effects of Limb-Length Discrepancy on Gait Economy and Lower-Extremity Muscle Activity in Older Adults. The Journal of Bone & Joint Surgery. 2001;83(6):907-915.
26. Woo RY, Morrey BF. Dislocations after total hip arthroplasty. The Journal of Bone & Joint Surgery. 1982;64(9):1295-1306.
27. Budenberg S, Redmond A, White D, et al. Contact surface motion paths associated with leg length inequality following unilateral total hip replacement. Proceedings of the Institution of Mechanical Engineers. Part H, Journal of engineering in medicine. 2012;226(12):968-974.

28. McWilliams AB, Grainger AJ, O'Connor PJ, Redmond AC, Stewart TD, Stone MH. Assessing reproducibility for radiographic measurement of leg length inequality after total hip replacement. Hip international : the journal of clinical and experimental research on hip pathology and therapy. 2012;22(5):539-544.