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2 Kinematics of the Native Knee

Ryan J. Reynolds, Aude Michelet, Jacobus H. Müller, and Mo Saffarini

Keynotes

1. Kinematics is a branch of physics concerned with analysis of movements, in absolute or relative space, without consideration for their driving or resistant forces.
2. Knee kinematics are determined primarily by the four ligaments, the ACL, PCL, MCL, and LCL.
3. Over the past two decades, various authors reported disparate kinematic patterns, which could be attributed to the heterogeneity of knee specimens, imaging modalities, reference axes, and loading conditions.
4. The tibiofemoral joint is a bicondylar, modified-hinge joint that also exhibits rotational and linear movements, thereby allowing up to six degrees of freedom during dynamic activities. The center of the rotation is located in the medial tibiofemoral compartment.
5. The main biomechanical function of the patella is to improve quadriceps efficiency by increasing the lever arm of the extensor mechanism.
6. The knee is considered to be stable when, in response to external forces, there are no subjectively excessive rotations or displacements, and the surrounding ligaments are within their elastic ranges.
7. The extent of knee flexion required for different activities varies considerably: 67° for walking, 83° when climbing stairs, 90° when sitting down and descending stairs, 106° when tying shoelaces, and 130° when squatting.
8. A clear understanding of the interrelationship between the different structures of the native knee joint and their role in knee kinematics is required to better serve the functional needs of patients.

2.1 Introduction

The knee serves several important functions, including sustainment of body weight, transmission of forces for motion, and conservation of momentum during gait [1]. Yet the knee is the least stable joint in relation to the loads it supports [2]. Its intrinsic susceptibility to damage is mainly due to poor congruence between its

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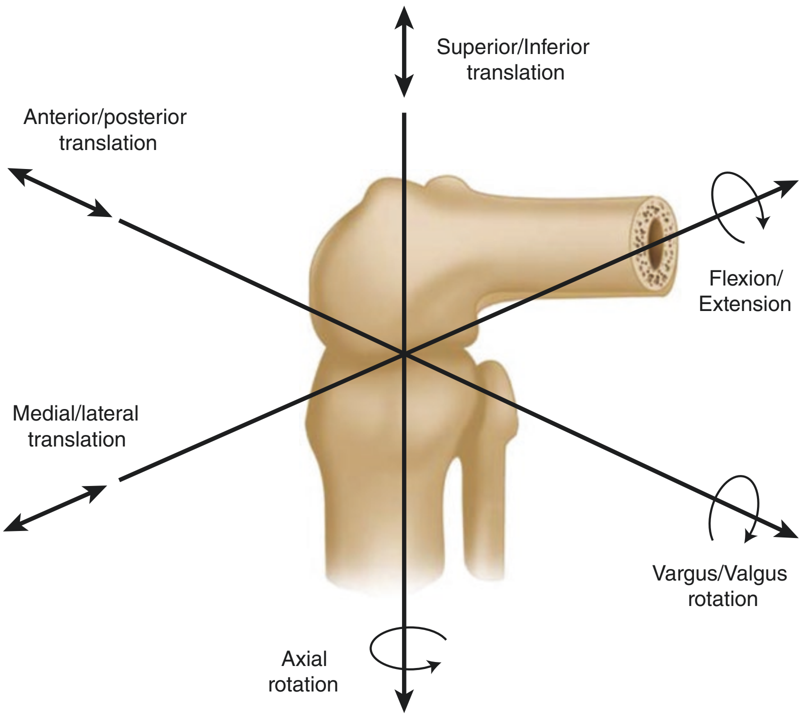

Fig. 2.1 Six degrees of freedom of the knee joint

articular surfaces and partly because of substantial dependence on its surrounding soft tissues for coherence [3].

Kinematics is a branch of physics concerned with analysis of movements, in absolute or relative space, without consideration for their driving or resistant forces [3]. In its simplest form, the knee can be represented as a simple hinge that allows pure flexion and extension about a single mediolateral axis, hence limited to one degree of freedom [4]. In reality, however, the knee is a bicondylar, modified-hinge joint that also exhibits rotational and linear movements, thereby allowing up to six degrees of freedom during dynamic activities: three rotations (flexion–extension, external–internal, varus–valgus) and three translations (anteroposterior, mediolateral, and compression–distraction) (Fig. 2.1) [1, 5].

Side Summary

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Understanding knee kinematics is of paramount importance to clinicians and surgeons, not only to enable them to restore normal function in pathologic or injured knees, but also to help diagnose and understand knee pathologies and injuries [3]. Knowledge of knee kinematics is equally important to biomedical engineers and sports scientists, particularly those involved in design or assessment of surgical implants and techniques for ligament reconstruction, meniscal repair, bone deformity correction, as well as partial or total arthroplasty [6].

In this chapter, the authors analyze kinematics of the native knee from various perspectives, starting with some reminders of the anatomic structures and articular geometries that guide the movements, followed by detailed representations of the physiologic patterns during different activities, and ending with a review of kinematic discrepancies between individuals, genders, age groups, and ethnicities. The authors attempt to include a balance of simple and complex analyses, to cover both historical and recent literature, and to explain patterns in clear and concise terms. Throughout the chapter, the reader should remember that the distinct kinematic patterns described within the knee are interdependent and are closely related to motions and loading of the adjacent joints of the lower limb, especially the hip and ankle.

Side Summary
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2.2 Physiology

The knee joint consists of four bones and three articular compartments [7]: (a) the medial tibiofemoral compartment (the medial condyle of the femur and the medial side of the tibial plateau), (b) the lateral tibiofemoral compartment (the lateral condyle of the femur and the lateral side of the tibial plateau), and (c) the patellofemoral compartment (the dorsal side of the patella and the femoral trochlea, extending to the distal condyles) [7]. The lateral tibiofemoral compartment is less stable than the medial tibiofemoral compartment, but it has greater mobility that serves to increase the range of motion of the knee and allow for internal–external rotation [3]. The articular surfaces of both medial and lateral tibiofemoral compartments are incongruent; their contact areas are thus limited and change through flexion. The meniscus increases the tibiofemoral articular contact area, thereby lowering contact pressure and improving the knee's congruence [3]. The patellofemoral compartment is incongruent with the medial and lateral tibiofemoral compartments when the knee is extended but becomes more congruent as the patella engages within the trochlear groove and begins to transmit loads beyond the first 20° of flexion [8].

The knee comprises four ligaments which help ensure knee stability through their viscoelastic properties and proprioceptive stress functions, thus preventing joint injury [3, 9]. The ligaments include (a) the anterior cruciate ligament (ACL), (b) the posterior cruciate ligament (PCL), (c) the medial collateral ligament (MCL), and (d) the lateral collateral ligament (LCL). The ACL originates from the inter-condylar notch of the femur and inserts slightly anteriorly in the center of the tibial plateau. Its primary function is to prevent excessive anterior translation of the tibia [10]. The PCL also originates from the inter-condylar notch of the femur and inserts posteriorly in the center of the tibial plateau. Its primary function is to induce femoral roll-back during knee flexion and thus increase range of motion. The PCL also restrains posterior translation of the tibia especially without load-bearing [11]. Without load-bearing, the ACL resists 86% of the anterior directed force, while the PCL resists 95% of the posterior directed force [10]. The MCL connects the medial margins of the femur and tibia, while the LCL connects the lateral margins of the femur and fibula. The MCL and LCL also contribute valgus–varus torsional stability, together with the joint capsule [12, 13]. A number of tendons (gastrocnemius, hamstrings tendon, patellar tendon, etc.) attach the flexor and extensor muscles, which therefore control knee motions and provide dynamic stability [14]. Beyond external loads and muscle forces, the geometry of the knee's articular surfaces, together with the configuration of its tendons and ligaments, are the chief determinant of knee kinematics. Even the slightest disruption or deformation to any of these anatomic structures could lead to abnormal kinematics that may prevent the individual from performing basic functions or cause further damage or injury [2].

Side Summary

The knee joint comprises three compartments: (a) the medial tibiofemoral compartment (the medial condyle of the femur and the medial side of the tibial plateau), (b) the lateral tibiofemoral compartment (the lateral condyle of the femur and the lateral side of the tibial plateau), and (c) the patellofemoral compartment (the dorsal side of the patella and the femoral trochlea, extending to the distal condyles).

Side Summary

The ACL and PCL together constitute a four-link bar in the knee [15]. The elastic flexibility of the ligaments functions as proprioceptive stress transducers, which help prevent joint injury [2]. Beyond external loads and muscle forces, the geometry of the knee's articular surfaces, together with the configuration of its tendons and ligaments, is the chief determinants of knee kinematics.

Unlike the ankle and wrist joints, which allow considerable rotation about both the anteroposterior (AP) axis (inversion and eversion) and the mediolateral (ML) axis (dorsiflexion/plantarflexion), or the hip and shoulder joints, which allow free rotation about all three axes (abduction/adduction, flexion/extension, and internal/external rotation), the primary kinematic functions of the knee and elbow joints are limited to rotation about the mediolateral axis (flexion–extension) [1, 3]. This over-simplified analogy must not detract from the importance of the auxiliary rotational and linear motions within the knee, which serve to stabilize it under different loading scenarios and to maximize its range of motion when needed.

The bipedal posture of humans doubles the loads borne by the knees and destabilizes them substantially compared to quadrupedal animals [1]. The knee is therefore highly susceptible to ACL injury if the femur and tibia are subject to opposing forces or moments, causing excessive varus–valgus, internal–external rotations, or even anteroposterior translation [9, 16]. Nevertheless, constant muscular reflexes and ligament tensions compensate for its inherent instability and often prevent falls and dislocations [2]. It has in fact been shown that neuromuscular training can reduce these risks and enables the joint to move with increased stability, even when non-muscular anatomic structures are unable to [16, 17].

The kinematics of the knee can be divided into tibiofemoral (TF) kinematics (grouping both the medial and lateral compartments) and patellofemoral (PF) kinematics [3]. The former is well studied and documented in orthopedic and sports medicine literature [18–28]. Although a series of in vivo and in vitro studies have been conducted on the latter [29–37], PF kinematics are somewhat less understood, with inconsistent descriptions [38]. Interestingly, the TF and PF joints exhibit different extents of rotational laxity depending on the knee flexion angle, and both joints lock their rotational positions to grant stability when needed. The TF joint locks in a rigid rotational position between full extension and 10° of flexion, but gains considerable rotational laxity (femur rotates externally) between 30° and

140° of flexion [39]. The PF joint is conversely lax between full extension and 20° of flexion, but the patella locks securely within the trochlear groove between 30° and 140° of flexion [40]. What might seem a coincidental reversal of rotational locking versus laxity, between 20° and 30° of flexion, is an important aspect in knee physiology, crucial to preventing subluxations or dislocations between different bones [3].

Side Summary
Knee kinematics can be divided into kinematics of the medial and lateral tibiofemoral compartment and the patellofemoral compartment.

Side Summary
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2.3 The Lower Limb Kinetic Chain

Before studying kinematics of the TF and PF joints in detail, it is important to understand the principal loading conditions in the knee joint and to consider that the weight and motion of the body are supported and governed by the entire lower limb, of which the knee is only one of several articular joints.

The knee joint supports the body by distributing its weight over the medial and lateral TF compartments. The contact stresses in these compartments are attenuated by the menisci, which help distribute loads more evenly over a greater surface area [1, 7]. Furthermore, the anterior tibial slope and the menisci also contribute to the anteroposterior stability of the knee, although knee stability depends mostly on the soft tissues (ligaments and tendons with their respective muscles) surrounding the joint [7, 41].

It is important to note that the knee is an integral part of the body's kinetic chain which, comprised of the spine, hips, knees, and ankles, controls lower extremity movements [16, 42]. The kinetic chain model refers to the body as a linked system of interdependent segments, often working in a proximal-to-distal sequence, to achieve the desired movement in an efficient manner [43]. The proximal and distal segments of the kinetic chain have considerable effects on knee kinematics [44, 45] though these considerations will not be addressed here.

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2.3.1 Tibiofemoral Kinematics

The knee moves primarily as a hinge that closes (flexion) with the contraction of the hamstrings and opens (extension) with the contraction of the quadriceps. During flexion and extension, the femoral condyles glide and roll over the tibial plateau [1, 5]. The extent of rotational and linear movement is governed by contractions of the hamstring and quadriceps muscles, and restricted by tensions within the ACL and PCL at different flexion angles [46, 47]. The posterior translation of the femur relative to the tibia or anterior translation of the tibia relative to the femur during flexion, known as *"femoral roll-back"* and "*tibial roll-forward,"* respectively, are most pronounced during midflexion (30° to 120°), and are crucial to enable deep flexion (beyond 120°) [1, 7]. Moreover, condylar asymmetry causes more roll within the lateral compartment and more glide within the medial compartment, which leads to internal–external rotation within the TF joint. The external rotation of the tibia relative to the femur as the knee extends from 30° flexion to terminal extension—also termed the "*screw home mechanism*"—contributes to the aforementioned locking of the femur and tibia in extension [1, 3, 7].

The first study of knee kinematics dates back to the early nineteenth century, whereby Weber and Weber [48] made direct visual observations on cadaveric specimens and described the medial motion of the femur onto the tibial plateau to be "cradle-like." Since then, several authors confirmed these observations using quantitative in vitro cadaver studies as well as in vivo imaging analyses. The advancement of computed tomography (CT) and magnetic resonance imaging (MRI) later enabled quantification of tibiofemoral displacements at different flexion angles and in different loading scenarios [21, 25, 27, 49, 50]. Most recent studies of TF kinematics illustrate the relative positions of the femur and tibia using two-dimensional (2D) coordinates in the sagittal [51–55] and transverse planes [18, 19, 49, 56–59].

Side Summary

The posterior translation of the femur relative to the tibia during flexion is known as *"femoral roll-back" (*or *"tibial roll-forward")*. It is most pronounced during mid-flexion (30°–120°) and is crucial to enable deep flexion (beyond 120°). The external rotation of the tibia relative to the femur as the knee extends from 30° flexion to terminal extension is termed the "*screw home mechanism*" and contributes to the aforementioned locking of the femur and tibia in extension.

2.3.1.1 Sagittal Plane

Sagittal plane representations help visualize the knee in various flexion angles, including femoral roll-back and patellar position, but do not illustrate the screw home mechanism. This view enables analysis of the flexion–extension motion of the knee, which is often divided into three arcs: (a) the "screw home arc" (0°–30°), (b) the "functional arc" (30°–120°), and (c) the "passive arc" (120°–160°), with 0° corresponding to full extension (Fig. 2.2) [60]. The screw home arc is thus termed due to the marked rotation of the femur relative to the tibia as the knee approaches full extension: The lateral femoral condyle continues

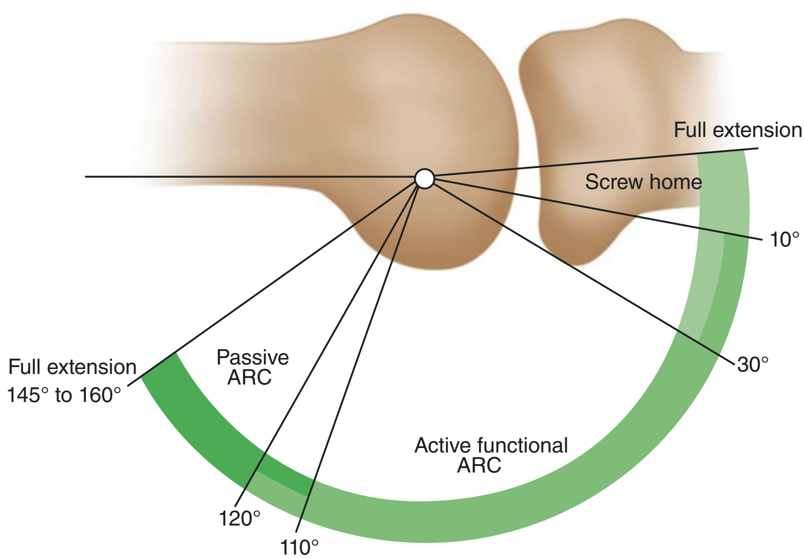

Fig. 2.2 The three arcs of the flexion–extension motion

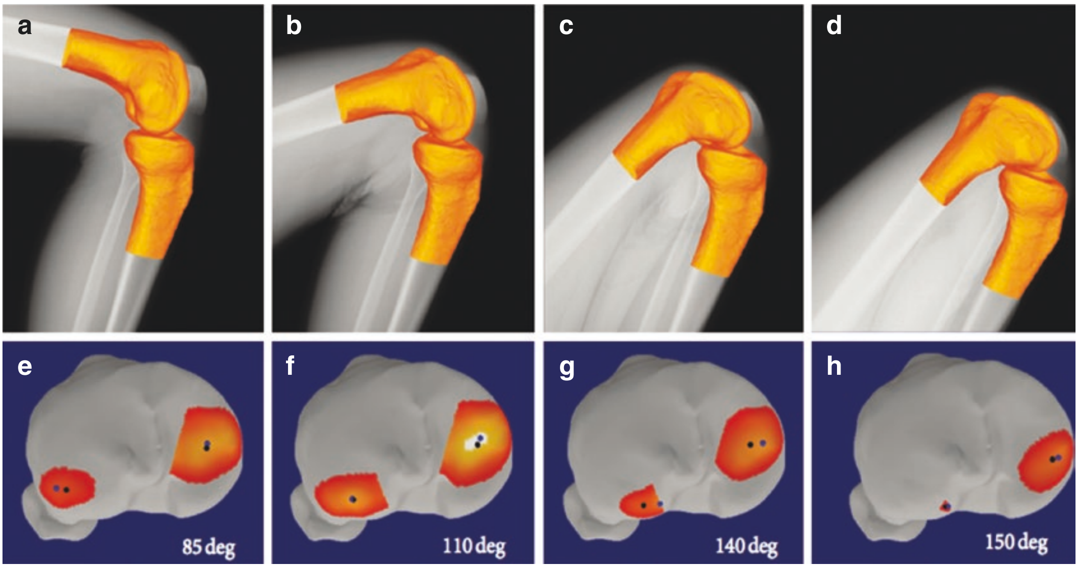

Fig. 2.3 Tibiofemoral contact areas at different flexion angles. (**a** and **e**) TF contact pattern at 85° flexion; (**b** and **f**) TF contact pattern at 110° flexion; (**c** and **g**) TF contact pattern at 140° flexion; (**d** and **h**) TF contact pattern at 150° flexion) (Adopted from Hamai et al. [41])

to translate anteriorly, while the medial femoral condyle exhibits minimal anterior displacement, thereby acting as a "medial pivot." The functional arc is the range where muscle activity and joint reaction forces are greatest: the femur continues to rotate relative to the tibia during flexion but at a much slower rate. The passive arc is so named as it cannot be reached through muscle contraction and instead requires body weight or an extrinsic force to induce flexion. At the more extreme end of flexion, the lateral side translates posteriorly to the point of subluxation (Fig. 2.3). Without this translation, deep flexion would be either impossible or painful. Frankel et al. [61] were among the first to describe the flexion–extension axis as a moving "instantaneous center of rotation." Using "true-lateral" X-rays, the authors showed how, on normal knees, the instantaneous center of rotation moves through a semi-circular pathway (Fig. 2.4) [1]. Several authors built on this model to determine precise locations of the flexion–extension axis at different angles [51–54]. The limitations of studies based on "instantaneous centers of rotation" include lack of a consistent Cartesian coordinate system, definition of the flexion–extension axis in two dimensions only, and inability to make continuous measurements.

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Sagittal plane representations enables analysis of the flexion–extension motion of the knee, which is often divided into three arcs: (a) the "screw home arc" (0°–30°), (b) the "functional arc" (30°–120°), and (c) the "passive arc" (120°–160°), with 0° corresponding to full extension. True-lateral X-rays reveal that the knee instantaneous center of rotation moves through a semi-circular pathway.

2.3.1.2 Transverse Plane

Transverse plane representations help to illustrate femoral roll-back and the screw home mechanism but require superimpositions of a line connecting the medial and lateral tibiofemoral contact points or projected centers of the femoral condyles on the surface of the tibia, plotted as a function of the flexion angle. This permits simultaneous visualization of femoral roll-back and screw home rotation during flexion (Fig. 2.5). Tanifugi et al. [62] reported that between full extension and 140° of flexion, the medial condyle translates over 20% along the tibial plateau (between 40% and 60% of the AP dimension), while the lateral condyle translates over 60% along the tibial plateau (30–90% of the AP dimension). Most other studies concur that knee flexion induces rotation of the tibia relative to the femur; in full extension, the tibia is externally rotated by up to 23°, while in full flexion, the tibia is internally rotated by up to 12° [18, 19, 49, 56–59]. They also agree that flexing the knee to 120° causes posterior translation of the lateral femoral condyle by up to 45 mm, and of the medial femoral condyle by up to 30 mm [19, 49, 56, 57, 59]. Despite considerable discrepancies, most authors agree that the medial femoral condyle has a relatively stable position [63]. By contrast, Feng et al. [64] and Pinskerova et al. [50] observed some initial anterior translation of the medial femoral condyle, followed by gradual posterior translation. In high flexion (>120°), Hamai et al. [41] reported a paradoxical "lateral pivot," while Johal et al. [22] emphasized that the medial and lateral condyles had equal posterior translations. The results are dependent on how the experiments were conducted, whether with or without axial loads, and how the knee was flexed, either passively or under quadriceps contraction. High flexion kinematics are also variable according to the activity that is performed [65].

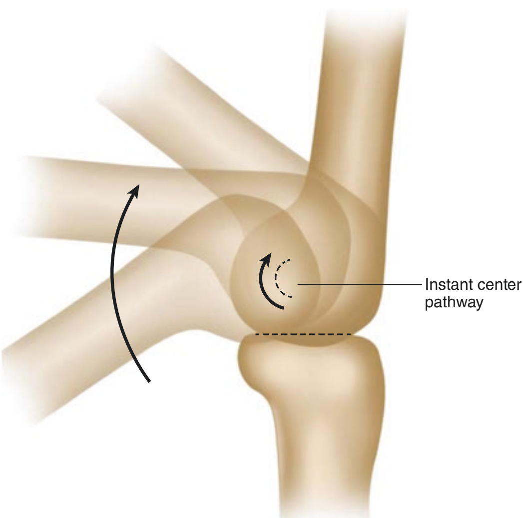
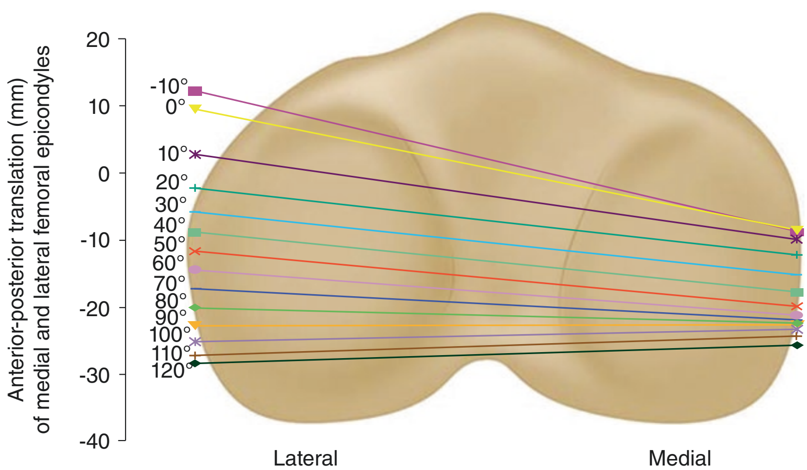


Fig. 2.4 The semi-circular pathway of the instantaneous center of rotation during knee flexion

Over the past two decades, various authors reported disparate kinematic patterns, which could be attributed to the heterogeneity of knee specimens, imaging modalities, reference axes, and loading conditions. On the one hand, in vitro cadaver studies enable fitting bones within sophisticated experimental rigs or optical trackers [21, 24, 46, 52, 66], which grant high accuracy. On the other, in vivo patient studies allow simulation of real loading with natural muscle contractions but require advanced imaging technologies [25, 49, 62, 64, 67]. Fluoroscopy enables real-time observation of in vivo knee kinematics [25, 26, 49] but does not reveal soft-tissue structures, while MRI provides excellent volumetric detail but is typically restricted to static analyses, with only a few studies describing methods for dynamic acquisition [68–70]. Even when taking measurements on the same specimen, using identical imaging techniques and loading

Fig. 2.5 Anteroposterior translation of the medial and lateral femoral epicondyles during knee flexion, projected in the transverse plane on the tibial reference system



conditions, a number of studies highlighted how the choice of reference axis could considerably alter findings [62, 64, 67]. For instance, Tanifugi et al. [62, 67] reported femoral rotation during flexion to be about 26° when using the geometric center axis (GCA), and about 17° when using the clinical transepicondylar axis (cTEA). They further showed that while GCA and cTEA offer approximately similar measurements on the lateral side, they differ significantly on the medial side because the two axes have different starting positions and paths during flexion (Fig. 2.6). Feng et al. [64] demonstrated similar findings but emphasized that on the medial side the use of the cTEA or GCA reveals some anterior translation of the medial condyle prior to its posterior translation. Victor et al. [71] illustrated the noticeable effect of contractions within the hamstrings and quadriceps on TF translations and rotations, which can be attenuated or reversed depending on loading conditions. The variability of TF kinematics depends on the flexibility allowed by the surrounding soft tissues, which provide multiple motion paths within certain boundaries.

Side Summary
Transverse plane representations help illustrate the femoral roll-back and screw home mechanism but require superimpositions of a line connecting the medial and lateral tibiofemoral contact points or projected centers of the femoral condyles on the surface of the tibia, plotted as a function of the flexion angle. Knee flexion induces rotation of the tibia relative to the femur; in full extension the tibia is externally rotated by up to 23°, while in full flexion, the tibia is internally rotated by up to 12°.

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Over the past two decades, various authors reported disparate kinematic patterns, which could be attributed to the heterogeneity of knee specimens, imaging modalities, reference axes, and loading conditions.

2.3.2 Patellofemoral Kinematics

The main biomechanical function of the patella is to improve quadriceps efficiency by increasing the lever arm of the extensor mechanism (Fig. 2.7) [72]. The patella does so by displacing the patellar tendon away from the tibiofemoral contact point, thereby increasing the mechanical advantage of the quadriceps during knee extension [73–79]. The position and orientation of the patella relative to the tibiofemoral joint determine the lever arm of the extensor mechanism and therefore influence required quadriceps forces [74, 78], joint reaction forces [75, 80, 81], and the level of contact with the femoral trochlea and condyles [82–84]. Patella tracking refers to the articulation pattern of the patella relative to the trochlear groove during knee flexion. Although the patella has six degrees of freedom, the patella tracking parameters of interest are patella shift, patella height, and patella tilt (Fig. 2.8) [85]. Consensus between studies reporting on patella tracking is largely affected by the inconsistent definitions of the applied coordinate systems, reference points, and the experimental protocols [38, 86].

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2.3.2.1 Patella Tracking

In full extension, the distal attachment of the patellar tendon on the tibial tubercle is positioned laterally in relation to the trochlear groove [8],

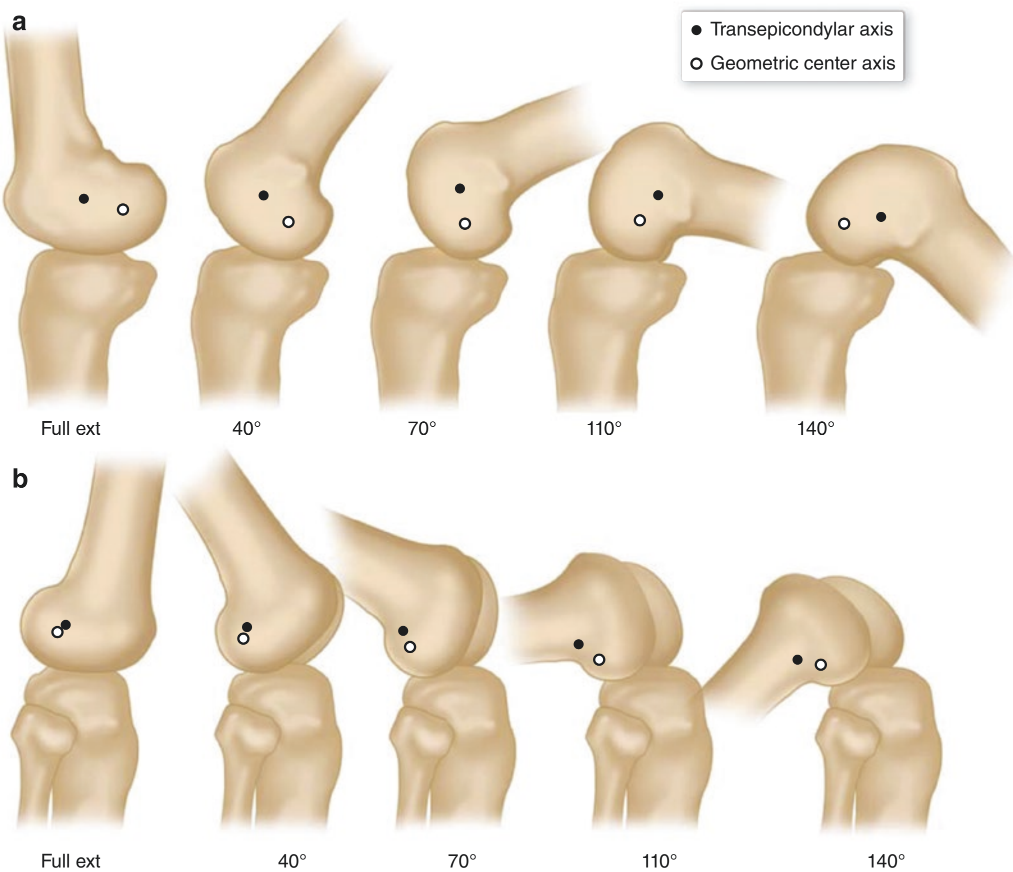

Fig. 2.6 Comparison of geometric center axis (GCA) and clinical transepicondylar axis (cTEA). (**a**) medial view; (**b**) lateral view

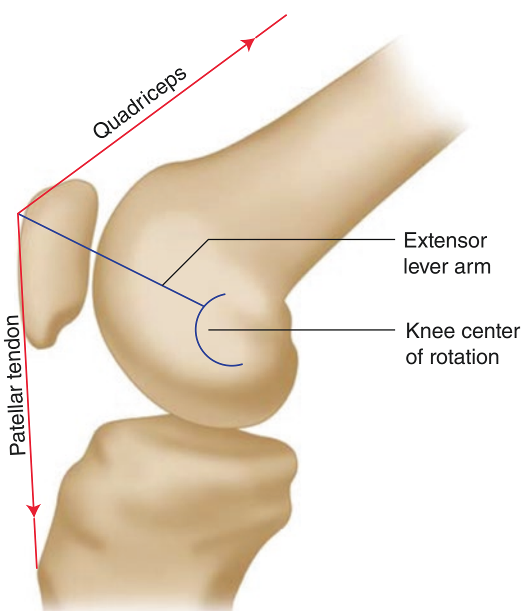

Fig. 2.7 The biomechanical advantage of increasing the extensor mechanism lever arm with the aid of the patella

and the patella is not congruent with the trochlear groove [38]. The angle forming between the effective quadriceps vector and patellar tendon vector is referred to as the Q-angle and leads to a lateral pull on the patella in full extension (Fig. 2.9). This lateral force is resisted by the oblique vastus medialis muscle, medial patellofemoral ligament, and the lateral trochlear facet. As the knee starts to flex, the tibia rotates internally relative to the femur, thereby decreasing the Q-angle, and the patella enters the trochlear groove from the lateral side [8].

After engagement with the trochlea, the patella will shift medially between 10° and 30° of knee flexion, after which it translates laterally again [38]. Some studies [87–89] indicate that the patella will shift medially beyond flexion angles of 80°, but there are limited data available beyond 90° of flexion since few studies consider deep flexion [88, 90]. There is no guidance on the clinical diagnosis and management for patellar proximal–distal and anterior–posterior displacement; hence, research on these two degrees of freedom is scarce [91]. Between full extension and 90° flexion, the patella will tilt medially between 1° and 3° and laterally between 1° and 15.5° [38]. During knee flexion, studies [88–90, 92, 93] indicate that the patella flexion angle will range between 60 and 70% that of the knee flexion angle [38]. The average curve derived from studies [92, 94–96] shows that the patella will rotate slightly medially at the beginning of flexion before its long-term lateral rotation with transient fluctuation [38].

Side Summary
As the knee starts to flex, the tibia rotates internally relative to the femur, thereby decreasing the Q-angle, and the patella enters the trochlear groove from the lateral side [8].

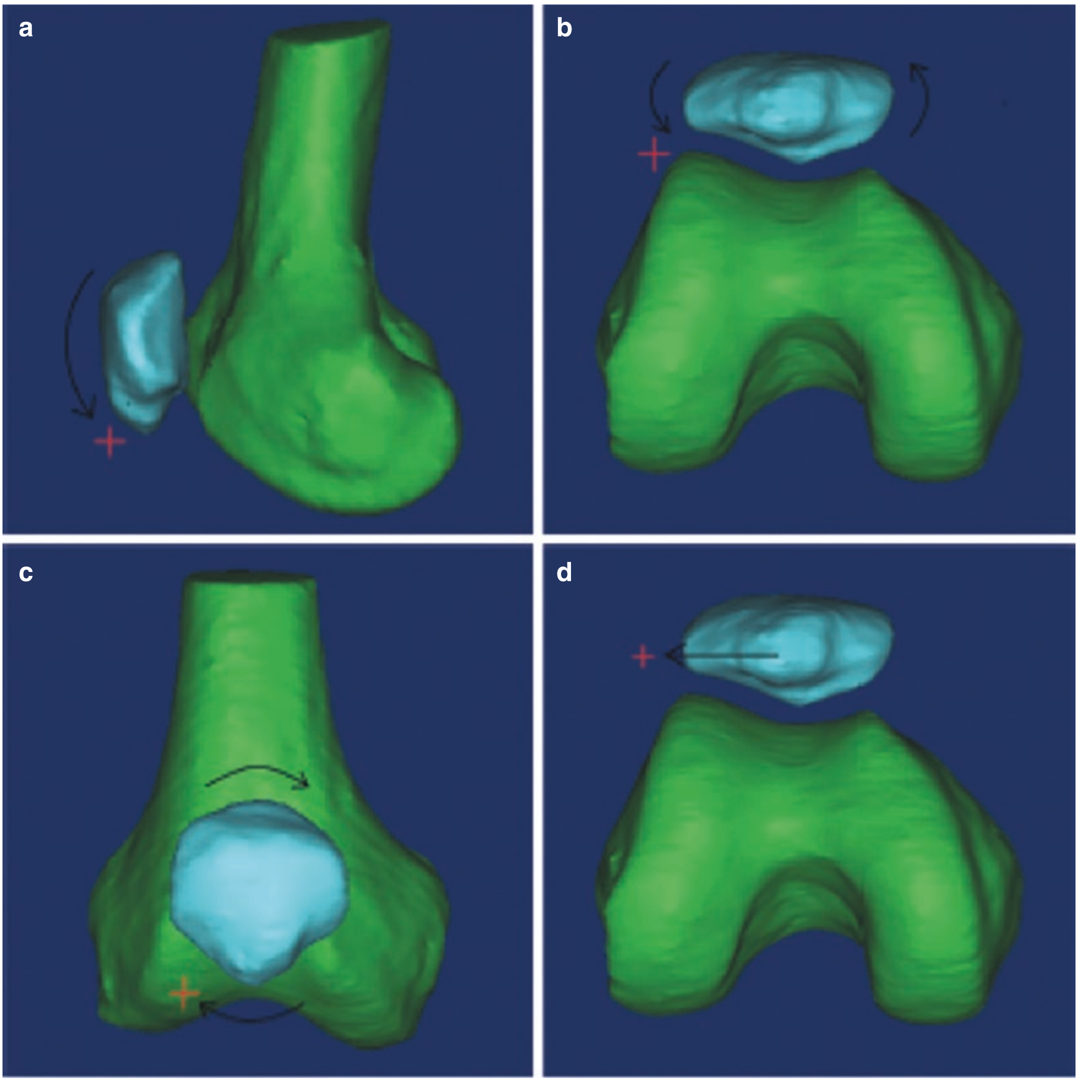

Fig. 2.8 Six degrees of freedom of the patella illustrated on a right knee joint. (**a**) Flexion–extension; (**b**) tilt; (**c**) rotation; (**d**) medial–lateral shift; (**e**) anterior–posterior translation; (**f**) proximal–distal translation (Adopted from Yu et al. [38])

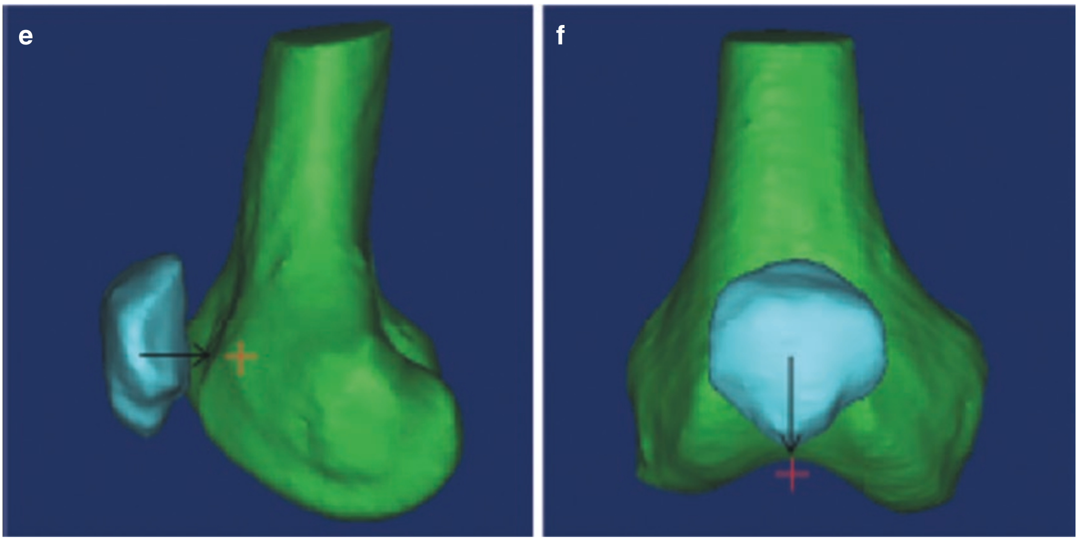

Fig. 2.8 (continued)

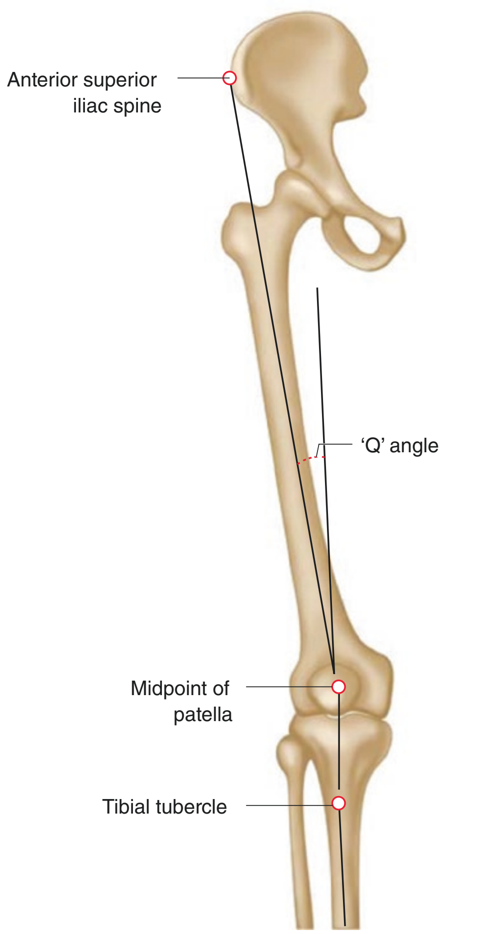

Fig. 2.9 Orientation of the effective quadriceps tendon and patella tendon force vectors to form the Q-angle

2.3.2.2 Patellar Height

The height of the patella relative to the trochlear groove is an important orthopedic measurement [85]. Although various methods to quantify patellar height have been proposed, there is no consensus in the literature on the most appropriate method or cut-off values [97]. The five most popular methods include the Insall–Salvati ratio [98], the Blackburn–Peel ratio [99], the Caton–Deschamps ratio [100], the modified Insall–Salvati ratio [101], and the Patellotrochlear index [102] (Fig. 2.10). In a recent comparison between the five methods, use of the Insall–Salvati ratio delivered better intra- and inter-observer reliability, whereas the use of radiographs and CT also provided better reliability in comparison to MRI [97].

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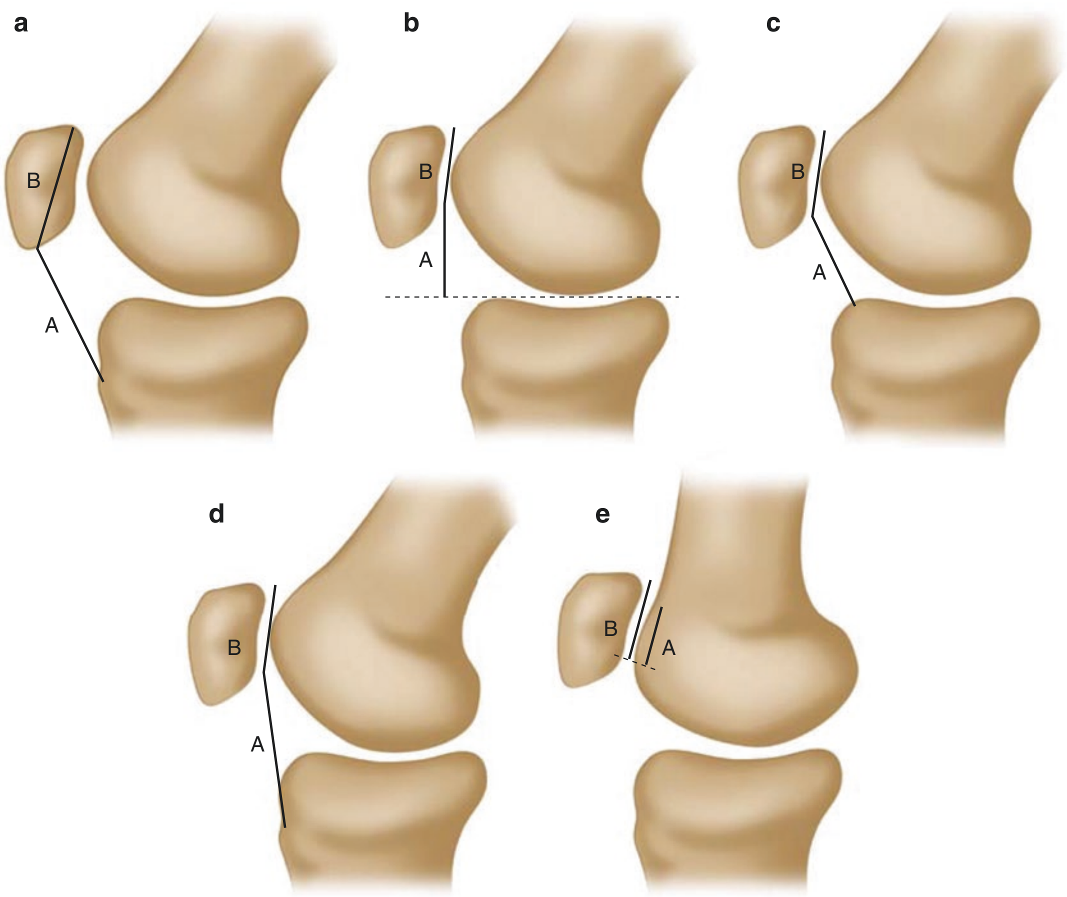

Fig. 2.10 The measurement of patella height = A/B. (**a**) Insall–Salvati ratio; (**b**) Blackburn–Peel ratio; (**c**) Caton–Deschamps ratio; (**d**) modified Insall–Salvati ratio; (**e**) Patellotrochlear index

2.3.2.3 Tibial Tubercle–Trochlear Groove Distance

The tibial tubercle–trochlear groove distance (TT-TG) is the measurement of the deepest point on the trochlear groove and central position of the patella tendon insertion on the tibial tubercle along the medial–lateral dimension (Fig. 2.11) [103]. Measurement of the TT-TG was originally defined using CT scans [104], but the use of MRI has also been described in the literature [103, 105, 106]. Although values reported in the literature show a high degree of variability [105], there is consensus that values exceeding 15 to 20 mm are pathological [103, 105]. It is known that the TT-TG will also vary between flexion angles and load-bearing conditions [105]. In a recent systematic review and meta-analysis comparing TT-TG measured with CT or MRI, the results indicated that TT-TG was a reliable measurement to differentiate between patients with and without patella instability [103]. TT-TG measured on CT was, however, significantly greater than the TT-TG measured on MRI, which suggest that different cut-off values should be used.

2.3.3 Stability

Due to the poor congruence of its articular surfaces, the knee is a relatively unstable joint in relation to the loads it supports [2]. Because it depends heavily on soft tissues to maintain

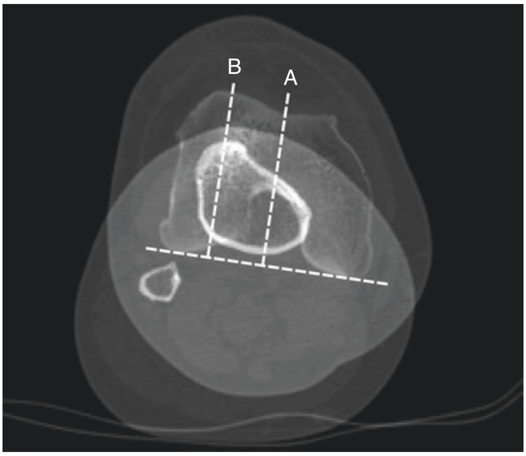

Fig. 2.11 The tibial tuberosity–trochlear groove (TT-TG) distance is measured using two superimposed CT slices: the first (A) through the most proximal part of the trochlear groove, where the notch looks like a Roman arch, and the other (B) through the most proximal part of the tibial tuberosity. The two reference points are projected perpendicularly to the bicondylar line. The distance between their projections is the TT-TG value

coherence, the knee is susceptible to injuries, particularly tears of the ACL [9, 16]. To prevent or repair injury, one must understand the mechanics of knee stability. In their seminal observational studies, Brantigan and Voshell [107] and Abbott et al. [108] introduced the general concepts of laxity and stability by describing the loosening and tightening of knee ligaments during flexion, their elongation when shear or torque loads were applied, and the effect of the interacting bearing surfaces on ligament lengths. The knee is considered to be stable when, in response to external forces, there are no subjectively excessive rotations or displacements and the surrounding ligaments are within their elastic ranges. Knee stability can be quantified in terms of knee laxity, evaluated by measuring the displacement (anterior–posterior, mediolateral, internal–external) or rotation relative to a neutral position when applying a force (or torque) to the femur or tibia. In a recent study, Marouane et al. [109] showed that the neutral position depends upon the posterior tibial slope and varies from one subject to another. The total laxity is determined by the net amount of displacement when applying a force in one direction and then applying the force in the opposite direction after returning to neutral.

Laboratory studies have focused on the primary roles of the different structures in providing stability. Girgis et al. [110] and Furman et al. [111] studied the anatomy of the cruciate ligaments to understand their ability to restrain anterior–posterior shear forces and identified two bands (or major fascicles) of each cruciate ligament, which loosened and tightened at different flexion angles. They used the method of selective resection of ligaments, which entails resecting one ligament at a time and testing the knee after each resection. By applying forces, they determined the relative contribution of each knee ligament to the general stability of the knee. Their study found that anterior translation increased most when the anteromedial band was severed, and further translation was seen with the severing of the posterolateral band and the medial collateral ligament (MCL). This study also highlighted that while the knee was in extension, the ACL limited both internal–external rotation and hyperextension. Finally, they found that during flexion there were fibers that stretched and contracted, and others that remained at constant length. These findings were confirmed by several other studies on knee stability, usually in the context of diagnosing soft-tissue injuries [10, 112–115].

A limitation with many of these early studies is that the knee was not axially loaded as it usually is in activity. Thereafter, Wang and Walker [116] showed that a compressive load substantially reduced rotary laxity and attributed to the geometrical interaction between the bearing surfaces. This work was followed up with a study of anterior–posterior and rotational laxity using selective cutting of ligaments and menisci to show their limited role in stabilizing the knee under load [117]. Knee stability under load was largely explained by the "uphill mechanism" where the femur would distract from the tibia in displacement or rotation. This is seen on the medial condyle as it has to climb out of the depression in the medial tibial compartment when experiencing shear forces, while the lateral condyle rests on the flat or convex surface of the lateral tibial compartment [118].

The reduction of laxity when the knee is loaded was confirmed in clinical studies. Markolf et al. [118] observed that AP laxity reduced by up to 50% when the patients tensed their muscles. Markolf et al. [119] later found that AP laxity reduced by only 30% in an unconstrained dissected cadaveric knee under load (925 N). These studies thus highlight the contribution of muscle contractions to knee stability, in addition to strains within the ACL [120], meniscus [13, 121], and cartilage [122].

Side Summary

The knee is considered to be stable when, in response to external forces, there are no subjectively excessive rotations or displacements, and the surrounding ligaments are within their elastic ranges. Knee stability can be quantified in terms of knee laxity, evaluated by measuring the displacement (anterior–posterior, mediolateral, internal–external) or rotation relative to a neutral position when applying a force (or torque) to the femur or tibia.

2.4 Kinematics during Different Activities

The extent of knee flexion required for different activities varies considerably: 67° for walking, 83° when climbing stairs, 90° when sitting down and descending stairs, 106° when tying shoelaces, and 130° when squatting. The loads transmitted through the knee at each flexion angle also vary depending on these activities, during which the native knee joint has variable degrees of congruency and stability [26]. A number of authors investigated how knee kinematics vary during different common activities. Their interesting observations are reported in the remainder of this section.

2.4.1 Walking

Walking, also termed "gait," has two principal phases: the stance phase and the swing phase. The stance phase is when the foot is on the ground, and the swing phase is when the foot is in the air. Each phase can be described in multiple parts. The stance phase includes initial contact, loading, mid-stance, terminal stance, and pre-swing. The swing phase includes initial swing, mid-swing, and terminal swing. The terminal swing ends with the initial contact portion of the stance phase. During the stance phase, the knee has a limited flexion of less than 10°, while during the swing phase, the knee flexes up to 55° (Fig. 2.12).

2.4.2 Stair Climbing and Descent

When climbing stairs, the knee has a maximum flexion ranging from 79° to 97°, a minimum flexion of 17°, and an internal rotation up to 15° [26, 123]. While descending stairs, the medial condyle translates anteriorly about 3 mm and the lateral condyle translates posteriorly approximately 7 mm [49]. The flexed knee and shifting body weight cause a slight paradoxical (anterior) motion on the medial side. Similar to gait, the majority of translation of the lateral condyle seemed to occur from heel strike to 66% of stance phase (average, 3.9 mm) as the lateral condyle moved in the posterior position [49].

2.4.3 Sitting Down and Standing from Seated

Sitting down has specific knee kinematics. The maximum flexion is slightly over 90° (94–97°), and the minimum angle is with the knee slightly

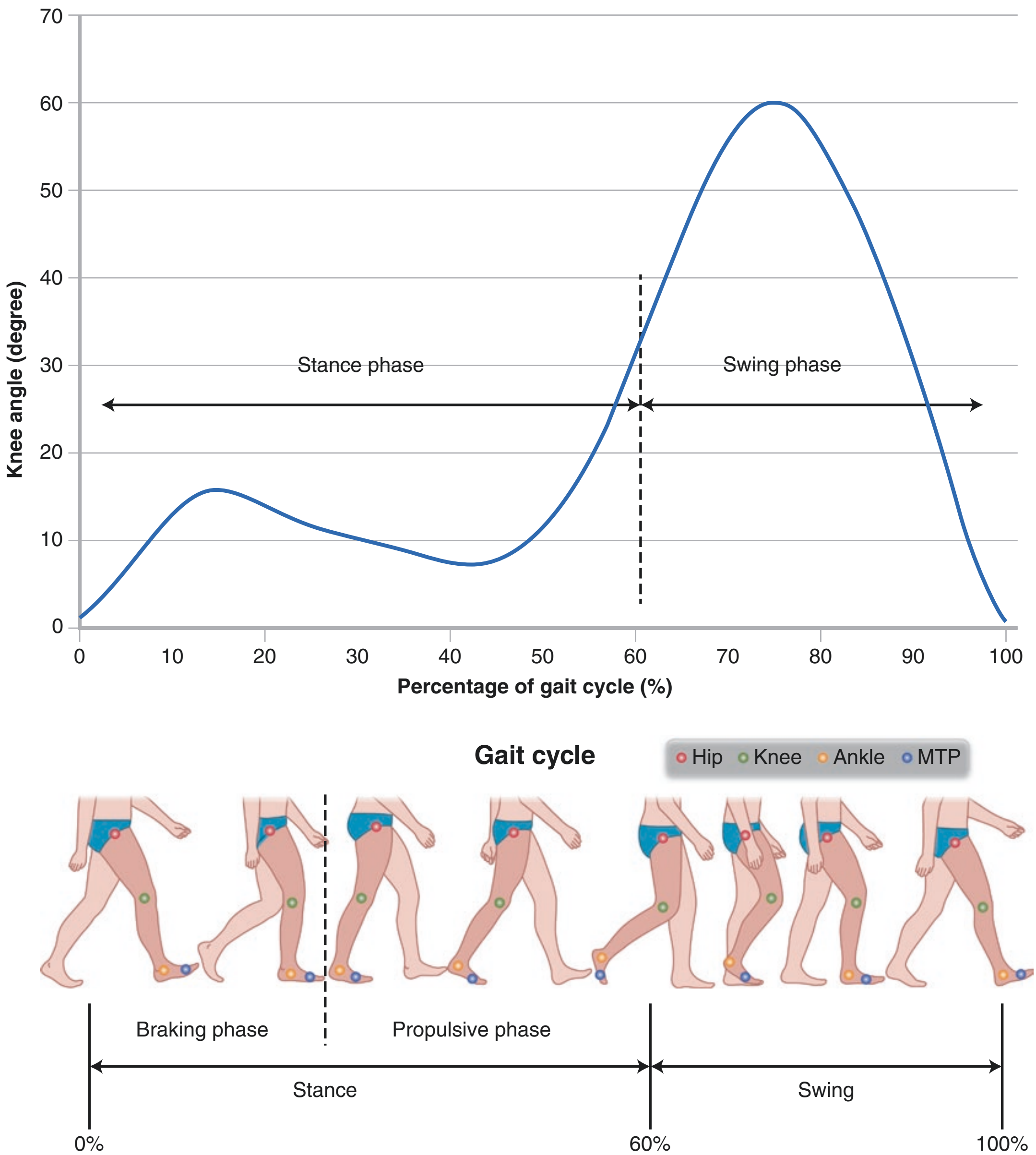

Fig. 2.12 Stance and swing phases of gait cycle

flexed (6°–8°) [49, 124]. Translation of the medial femoral condyles is greater while sitting into a chair (3 to −9 mm) than while rising up from a chair (0.5–5.9 mm). The decreased translation demonstrates the increased knee stability due to muscle action required to overcome gravity.

2.4.4 Squatting, Lunging, and Kneeling

During squatting, lunging, and kneeling, the knee flexion reaches its greatest extent [65]. Hamai et al. [125] had healthy individuals perform a lunge, enabling a single knee to be in view of the radiograph, flexing from the middle of the functional arc into the passive arc (85–150°). Their study evaluated the medial and lateral TF compartments, as well as femoral valgus rotation. Over the range of flexion, the medial side displaced anteriorly about 3 mm and then posteriorly about 4 mm, while the lateral side consistently displaced posteriorly about 8 mm. The knee externally rotated from 15° to 30° and moved from a slight varus rotation of 1° to a valgus rotation of 5°.

2.4.5 Vertical Drop Jump

One test of the ACL's condition is to perform a vertical drop jump (VDJ), where the subject jumps to the floor from a box 30 cm high. The medial–lateral motion of the knee is observed in assessing the status of the ACL. Krosshaug et al. [126] and Leppanen et al. [17] both detailed the VDJ, with Krosshaug et al. [126] evaluating a cohort of female handball and soccer players and Leppanen [17] evaluating both male and female floorball and basketball players. Krosshaug et al. [126] report that a VDJ was not able to establish risk of ACL injury, and that the only factor that was associated with risk of injury was medial knee displacement. Across all participants the average medial knee displacement observed for those that had a new ACL injury was 2.7 cm, while those that had no injury was 2.2 cm. The difference between other kinematic data was insignificant, and therefore one can expect to see about 2° valgus at initial contact and a peak knee flexion of 90° while performing a VDJ.

2.4.6 Sports

The majority of knee surgeries happen following sports injuries [127]. The knee and body go through more dynamic and aggressive motions than the controlled motions often reported. Steiner et al. [128] found that while 90 minutes of playing basketball or 10 km of running increased knee laxity by about 20%, squatting had almost no effect on AP laxity. Similarly, basketball players had greater valgus laxity after performing a jump landing compared to floorball players (−3 and −1 mm, respectively) [17].

Murakami et al. [28] evaluated the knee kinematics of five healthy males' golf swings, utilizing single-plane radiographs taken at 10 Hz. They found that the trailing knee rotated significantly more (26° on average) than the leading knee (18° on average) during a golf swing. Interestingly the external rotation of the left and right knee essentially mirrored each other; where there is external rotation of the left knee, the right knee will have internal rotation, and vice versa.

Side Summary
The extent of knee flexion required for different activities varies considerably: 67° for walking, 83° when climbing stairs, 90° when sitting down and descending stairs, 106° when tying shoelaces, and 130° when squatting.

2.5 Inter-Individual, Gender, Age, and Ethnic Variations

Komistek et al. [49] were among the first to highlight remarkable inter-individual variability of AP translations on the medial and lateral femoral condyles during flexion. Since then, numerous studies have investigated potential variations in knee kinematics across sex, age groups, and ethnicities.

2.5.1 Sexual Variations

There is some controversy as to whether there are meaningful differences in knee kinematics between men and women [17, 22, 23, 28, 129]. Nevertheless, it is worth noting the established differences in lower limb kinematics and muscle control between the sexes [16]. For instance, Leppanen et al. [17] found that a greater proportion of men had better knee control (75%) than women (21%), regardless of their sports activities, and that men's knees exhibited peak knee varus of 3.4° while women's knees exhibited peak valgus of 7.5°. Sheu et al. [130] found in a study testing side-cutting manoeuvres that men had greater flexion than women when entering a cutting motion. This difference could explain the greater susceptibility of women to ACL injuries. Mendiguchia et al. [16] observed that when performing sports manoeuvres, women had increased hip adduction and internal rotation. It is important to note that knee kinematics do not depend on the knee joint exclusively but also on the kinetic chain that controls lower extremity movements together with the spine, hips, and ankles. Thus, understanding knee kinematics requires having a systemic view of the lower limb, taking into account proximal and distal factors to the knee joint. For instance, women's altered spine and hip flexion angles, more lateral spine displacement, and larger ranges of spine motion when compared to men help explain their increased risk of ACL injury relative to males [16].

Side Summary
There are established differences in lower limb kinematics and muscle control between the sexes [16].

2.5.2 Age Variations

Age increases the risk for developing osteoarthritis and lowers muscle strength, both of which alter knee kinematics [20, 131–133]. Moreover, the recommended treatment for osteoarthritis is often total knee arthroplasty (TKA), so that studies comparing the performance of healthy knees to TKA knee are especially relevant for elderly patients.

In essence, aging normally slows knee motion and positions the knee in slight varus, both of which factors result in more work being required from adjacent joints to accomplish a task. In a study on 22 patients aged between 21 and 75, Fukagawa et al. [20] found that valgus angle and squat time significantly increased with age, and maximum flexion occurred later in the gait cycle. Likewise, Hortobágyi et al. [131] reported that elderly patients (mean 77 years) did more hip-positive work and less ankle-positive work during gait.

Side Summary
Age increases the risk for developing osteoarthritis and lowers muscle strength, both of which alter knee kinematics [20, 131–133].

2.5.3 Ethnic Variations Differences

In a study of healthy individuals of Japanese and Caucasian origin, Leszko et al. [23] evaluated whether sex or ethnicity had a greater effect on knee kinematics. They found that Caucasian men were limited in their maximum flexion compared to Caucasian women (respectively, 146° versus 152°), while Japanese men and women had similar ranges (respectively, 151° versus 153°). The authors also found that Caucasian men had their knees positioned more posteriorly, and as a result underwent less internal–external rotation, than the three other groups. In another study comparing Chinese, Malay, and Indian patients requiring TKA, Siow et al. [134] found small but significant differences in each ethnicity's preoperative range of motion.

Take Home Message
A clear understanding of the interrelationship between the different structures of the native knee joint and their role in knee kinematics is to be recognized. It can be expected that new rehabilitation protocols, surgical techniques, and treatment regimens will be developed based on this

understanding, to better serve the functional needs of the patient. The hypothesis of better functionality through kinematic normality has still not been achieved, primarily due to inconsistencies in coordinate reference frames, differences in measurement techniques, and inconsistent experimental protocols. There is a need for more guidelines like the ISB recommendations for joint coordinate systems [135] to reduce variability between different studies on kinematics.

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