Chapter 9
Biomechanics of the Knee

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Abstract Composed of tibiofemoral and patellofemoral joints, the knee is a complex synovial joint in the human body, both in terms of anatomy and biomechanics. Common knee injuries include torn ligaments and tendons as well as menisci that could lead to cartilage degeneration and osteoarthritis. Therefore, a thorough understanding of the anatomy and biomechanical function of the knee is important for diagnosing any knee disorders, developing appropriate clinical treatments, and improving the design of joint replacement prostheses. In this chapter, the functional anatomy of the knee and the biomechanics and kinematics of both the tibiofemroal and patellofemoral joints will be reviewed. Then, the knee pathomechanics and corresponding clinical treatments will be presented and discussed.

Keywords Knee joint · Biomechanics · Implant

1 Functional Anatomy of the Knee Joint

The tibiofemoral joint and patellofemoral joint are the two articulating surfaces that constitute the knee joint and both are diarthrodial joints that consist of bones, hyaline cartilage, ligaments, tendons, as well as the musculature (Fig. 9.1). The tibiofemoral joint also has medial and lateral menisci between its surface to distribute

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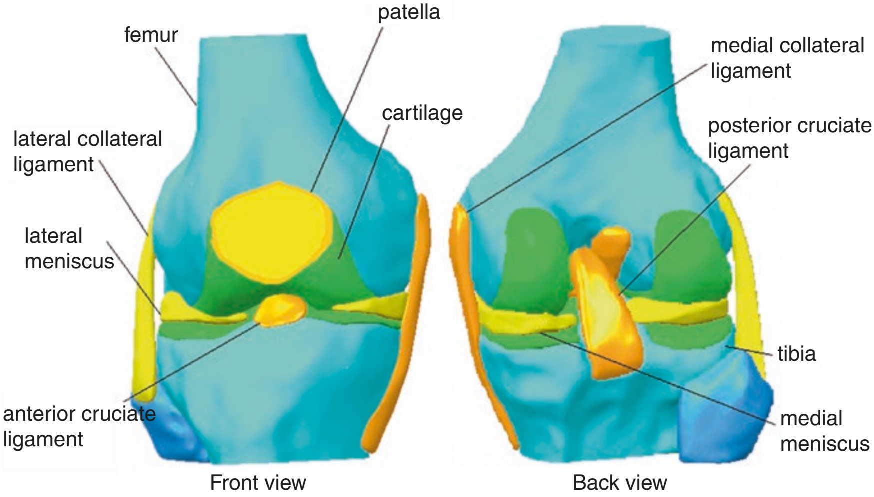


Fig. 9.1 Anatomy of the knee

the loads as well as to absorb impact forces on the joint. All these soft tissues work in synergy to stabilize the knee while the muscles power its motion.

Using the principles of biomechanics, the mechanical behavior and the function of the biological system, in this case the knee joint, can be analyzed and understood. The motion between two bony structures in a joint is known as kinematics. Normal kinematics of the knee is maintained by biomechanical functions of each joint component. Changing of any one of these components secondary to injury or diseases could alter the normal balance and synergy of the knee joint and introduce abnormal kinematics that could cause increased loading of other soft tissue structures and induce additional damages.

In this section, the anatomy and function of the knee components, as well as the joint kinematics, are presented.

1.1 Bones

As the femoral condyles are round while the tibial plateau is relatively flat, the conformity is reinforced by the menisci lying between them. In the sagittal view, the anterior section of the tibia is generally higher than its posterior section. Hashemi et al. [1] demonstrated that the posterior slope of the tibia ranged from $-3°$ to $14°$, and was steeper in females than the males. Also, in the coronal view, the tibial plateau is oriented upward in a medial-to-lateral direction. The coronal tibial slope ranged from $-1°$ to $6°$ and was less steep in females. Besides, there is typically a valgus angle of 7–10° between the tibia and femur. The medial condyle of the distal femur projects more distally than the lateral condyle [2].

The patella is a sesamoid bone and has the appearance of an inverted triangle. The cartilage of the patella is also the thickest in the human body [3].

1.2 Hyaline Cartilage

The hyaline cartilage in the knee joint is a layer of elastic tissue which covers the contact surfaces of the bones along which the joint moves. Hyaline cartilage is mainly composed of a matrix of proteoglycans and collagen traversed with interstitial water. This cartilage layer serves numerous functions, including providing a smooth surface for joint movement, buffering compressive loads, and protecting the underlying bone. With a thickness of between 1.69 and 2.55 mm, the hyaline cartilage in the knee is significantly thicker than in the hip (1.35–2.00 mm) and ankle joints (1.00–1.62 mm) [4].

The micro-structure of hyaline cartilage has low permeability, which acts to trap water within the matrix during loading. This pressurizes the cartilage and increases the stiffness by up to ten times that of the intrinsic modulus of the solid matrix [5]. The increased stiffness of the matrix means there is very little deformation, and the cartilage can take up to 90 min to recover after loading [6]. It has been shown that the thickest regions of the knee cartilage on the femur and tibia are aligned at full knee extension, suggesting an adaption of the cartilage to high loading such as during the heel strike phase of gait [7]. In contrast, under conditions of reduced loading, such as immobilization following surgery, the hyaline cartilage suffers atrophy. However, a sudden change in the loading pattern on the knee (for example, a knee with a ruptured ACL could cause a redistribution of loading on the articular surface) does not produce a significant increase in cartilage thickness, but may in fact initiate degeneration of the cartilage [7]. It has been reported that regular cyclic loading enhances the synthesis of proteoglycans which makes the cartilage stiffer, but continuous compressive loading hinders the synthesis of proteoglycans, which causes damage to the cartilage through necrosis [5].

1.3 Menisci

The crescent-shaped menisci are fibrocartilaginous structures lying between the contact surfaces of the tibiofemoral joint, plugging the gap between the femur and the tibia. To conform to the shape of the femoral condyle and the tibial plateau, the menisci are wedge-shaped in the coronal view, and have a concave upper surface that allows them to “hold” or stabilize the convex femoral condyles. The medial meniscus has a longer span in the A-P direction, while the lateral meniscus has a more rounded shape (Fig. 9.2). The menisci are connected to the tibia primarily by ligaments located at the four horns of the menisci, which restrict their movement relative to the tibial plateau but do allow some limited movement.

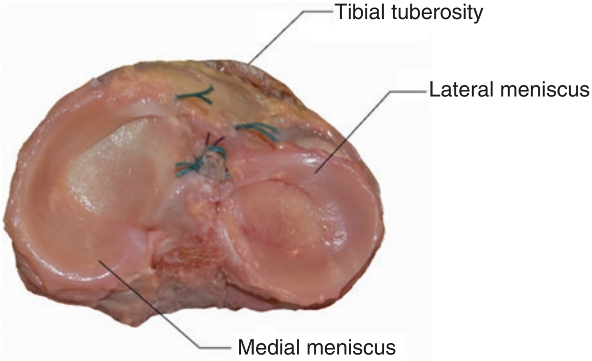


Fig. 9.2 Anatomy of the menisci

As the knee moves, the menisci deform to share and buffer forces in the joint and indirectly increase the articular contact area, thus stabilizing the knee and decreasing stresses on the hyaline cartilage of the tibia and femur. Studies have demonstrated that up to 70% of the loading on the lateral compartment of the knee is carried by the lateral meniscus while 50% of the loading on the medial compartment is borne by the medial meniscus [8].

The menisci also work with the hyaline cartilage to lubricate the knee joint and allow smooth motion between the articular contact surfaces. McCann et al. conducted cadaveric testing on bovine knees and demonstrated that removal of the menisci resulted in an increased friction coefficient on the joint [9].

Undue femoral gliding on the tibia plateau is also restrained by the menisci, making them particularly important for a knee with ACL deficiency. Previous studies have shown greater loading on the menisci following resection of the ACL [10].

1.4 Ligaments

Four major ligaments in the tibiofemoral joint are: the anterior and posterior cruciate ligaments (ACL and PCL), and the medial and lateral collateral ligaments (MCL and LCL). The ACL and PCL are intra-articular ligaments while the MCL and LCL are extra-articular (Fig. 9.1).

The bone insertions of the ACL are located at the medial posterior side of the lateral femoral condyle and the anterior side of the tibial plateau. Both insertion sites have a considerably larger cross section than the midsubstance of the ligament [11]. For biomechanical purposes, some researchers [11] have proposed dividing the ACL into two bundles according to the anatomical location of its tibial insertion: the anteromedial (AM) and posterolateral (PL) bundles. Length of the AM bundle is relatively constant through joint flexion angles from 0° to 90°, while the PL bundle is more stretched in extension [12]. However, pursuing this approach even further, other studies [13] have proposed dividing the ACL into three bundles (the anteromedial, intermediate, and posterolateral bundles).

The ACL plays an important role in constraining anterior tibial translation, thus preventing the femur from posterior luxation, while also helping to avoid excessive internal and valgus rotations of the tibia [14–16]. As the knee flexes, the femur initially rolls back along the tibial plateau during early flexion, before converting to anterior sliding due to the drag force from the ACL. Besides, previous studies [17] also showed an increased ACL force under a combined loading of axial compressive force and anterior tibial load, comparing with using an anterior tibial load alone, indicating a biomechanical role of ACL in resisting compressive joint impact.

The PCL inserts at the lateral posterior side of the medial condyle of the femur and stretches to the inferior posterior side of the tibial plateau. As with the ACL, the PCL may also be divided into two functional bundles, which are named according to the location of their femoral attachments [12]. Both the AL bundle and the PM bundle have increased lengths with flexion [12].

In contrast to the ACL, the PCL prevents the tibia from undue posterior translation, which decreases the risk of anterior luxation of the femur [18]. During knee extension, the PCL forces the femur to slide back along the tibial surface as it rolls forward. Besides, the PCL has been reported to prevent excessive external tibial rotation, especially for the joint flexion angles beyond 60° [19].

The MCL and LCL are located at the medial and lateral side of the knee, respectively. The LCL inserts at the lateral femoral condyle and the lateral superior side of the fibular head, while the MCL attaches to the medial femoral condyle and the inferior medial side of the tibial plateau.

The MCL and LCL function together to constrain the knee joint from excessive internal-external and varus-valgus rotations. The MCL cooperates with the ACL in constraining anterior tibial translation. This partnership is clearly evident in cases of ACL injury, which often result in a decrease in joint stability and an increase in the MCL force, subsequently placing the MCL at a greater risk of injury [20].

1.5 Muscle System

There are two major muscle groups primarily responsible for the extension and flexion of the knee: the quadriceps muscles, located at the front of the femur, which contract to extend the knee; and the hamstring muscles, located the back of the femur, which contract for knee flexion.

The quadriceps muscles consist of the vastus lateralis, rectus femoris, vastus intermedius, and vastus medialis muscles, all of which are controlled by the femoral nerve [21]. The rectus femoris muscle inserts at the hip bone and the tibia, while the other three muscles insert at the femur and connect to the quadriceps tendon.

The quadriceps muscles, mainly serving as a knee extensor, form a large fleshy mass which covers the front and sides of the femur. The rectus femoris muscle also acts as a flexor of the hip due to its attachment to the ilium [22]. As the knee extensor, the quadriceps muscles are crucial for most daily activities (walking, running, jumping, etc.) and serve to stabilize the knee. Weak quadriceps muscles are reported to compromise knee stability [23]. The lever arm of the quadriceps muscle is increased by the presence of the patella, which is discussed in more detail in Sect. 1.7.

The hamstring muscles consist of the semimembranosus, biceps femoris muscles, and semitendinosus, all of which are controlled by the sciatic nerve. The short head of the biceps femoris inserts at the trochanter of the femur and the lateral side of the fibular head, while the long head originates at the sciatic tuberosities and has the same fibular insertion point as the short head. The semitendinosus and semimembranosus muscles both originate from the sciatic tuberosities and insert at the medial side of the tibia.

The hamstring muscles act to flex the knee, while also function to extend the hip except for the short head of the biceps femoris [24]. The short head of the biceps femoris crosses only the knee joint and is therefore not involved in hip extension. The biceps femoris is an external rotator of the tibia, while the semitendinosus and semimembranosus muscles function to rotate the tibia internally [25]. As an antagonist to the quadriceps muscles, the hamstring muscles also function to control the joint motion during daily activities.

1.6 Biomechanics of the Tibiofemoral Joint

Motion and biomechanics of the tibiofemoral joint are greatly influenced by the morphology of the articular surfaces, and of course the integrity of all the joint structures [16]. The posterior tibial slope is considered an important characteristic for producing anterior shear forces on the tibia. Wang et al. suggested that in knee arthroplasty surgery, the posterior tibial slope has a significant effect on postoperative kinematics of the joint and thus the slope angle should be highly scrutinized intraoperatively [26]. Woo et al. found that increasing the posterior tibial slope caused an increased anterior tibial translation which could stabilize the knee with a deficient PCL. In contrast, decreasing the posterior slope may be considered for a knee with ACL deficiency [27]. Females have a deeper posterior tibial slope than males, resulting in a greater anterior tibial force, especially when in the posture of weight-bearing, placing females at greater risk of ACL rupture than males [1, 28]. Conversely, the lateral section of the tibial plateau is generally located higher (more proximal) than the medial section, forming a slope which is larger in males than in females, but the effect of this coronal slope on joint mechanics is not yet clear [1]. However, while the link between slop angle and knee mechanics has been widely studied, large differences between individuals means there is a large degree of variation across the general population [1]. Thus, understanding this variation could be beneficial for developing more accurate subject-specific implants that could better replicate normal knee biomechanics.

The tibiofemoral joint has 6 degrees of freedom (DOFs) of motion: three (anterior-posterior, proximal-distal, and medial-lateral) translations, as well as three (flexion-extension, varus-valgus, and internal-external) rotations (Fig. 9.3). Because the femoral

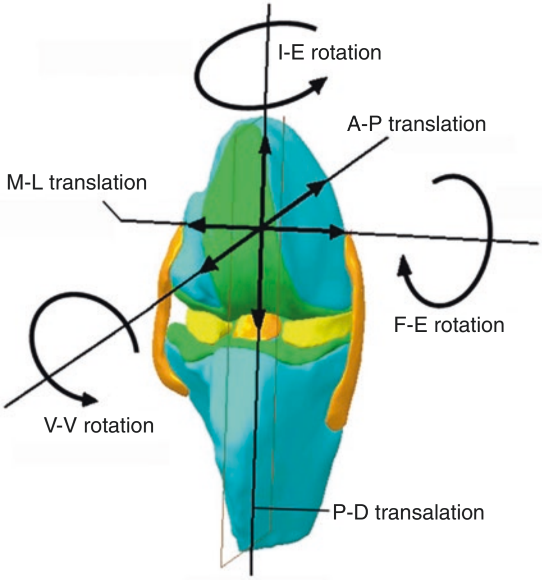


Fig. 9.3 Joint motions in 6 degrees of freedom

condyles have a circular shape and the tibial plateau is relatively flat, the tibiofemoral joint has good rotational flexibility in terms of flexion-extension (0–165°) [29].

The flexion movement of the knee joint can be divided into three functional phases, comprising the screw-home arc, the functional active arc, and the passive flexion arc [30]. The first phase is defined to represent joint activity through joint flexions from 0° to 20°, during which the joint motions are mainly determined by the morphology of the tibial plateau and the femoral condyles. Because of the asymmetry between the distal regions of the lateral and medial femoral condyles, the tibia undergoes internal rotation during this phase. The second phase represents joint activity from 20° to 120° of flexion, during which there is little axial rotation of the tibia. The final phase is defined as joint flexion over 120°. During this phase, the medial femoral condyle moves proximally because of its contact with the posterior section of the medial meniscus; this might explain why the posterior horn of the medial meniscus is typically ruptured during deep joint flexion [31]. Meanwhile, the lateral femoral condyle continues to move posteriorly during the passive flexion arc, resulting in a subluxation at the end of this phase.

During normal gait, the knee has a flexion angle of 0–10° at heel strike, then flexes to 15–20° at 15–20% of the gait cycle, followed by an extension during 20–40% of the gait cycle, and then flexes to around 60° in the swing phase [32]. Meanwhile, the knee undergoes internal-external rotation of up to 5° and varus-valgus rotation of up to 4°, combined with medial-lateral translation of up to 12 mm, proximal-distal translation of up to 14 mm, and anterior-posterior tibial translation of up to 7 mm [33]. As the knee is flexed, the medial condyle of the femur has little anterior-posterior movement but the lateral condyle rolls backward on the tibia, resulting in a coupled internal tibial rotation [34].

During normal gait, axial compression on the articular contact surface of the knee ranges from 0 to 3.2 times body weight (BW) [35]. The greatest axial compression occurs at around 50% of the gait cycle, just before the toe-off phase, and most of the joint forces are distributed on the medial tibial plateau. Loading in the A-P and M-L directions is much lower than the axial loading, ranging from 0 to 0.28 BW and 0 to 0.14 BW, respectively. Internal-external moments range from 0 to 0.02 Nm per BW.

1.7 Biomechanics of the Patellofemoral (PF) Joint

As the name suggests, the PF joint mainly consists of the patella and the femur. The quadriceps tendon inserts at proximal patella which connects the patella to quadriceps femoris muscle. On the distal side, the patellar tendon attaches directly to the tibial bone. As a gliding joint, the patella slides along the femoral trochlea during knee flexion/extension. The primary function of the PF joint is to increase the lever arm of the quadriceps muscles and thus to enhance the efficiency of extending the knee [36].

During flexion, the patella moves posteriorly and distally with respect to the femur condyles, following a “C”–shaped pattern (Fig. 9.4). Movement of the patella outside the sagittal plane is minor by comparison, tilting and twisting by less than 15° [37]. The lateral facet of the patella has been reported to bear greater contact forces than the medial facet during knee flexion [37]. Contact between the articular surfaces of the PF joint reach a peak when the joint flexes to 90°. Past 90°, the patella starts to rotate laterally and contacts the femur at the medial side. During deep flexion (around 140°), the patella falls into the femoral notch and the contact area is greatly reduced (termed the “odd facet”), resulting in high localized stresses. Therefore, patellar chondromalacia and patellofemoral pain syndrome are often experienced during this “odd facet” stage [38].

To have a better understanding on the biomechanics of the PF joint, the quadriceps angle (Q angle) needs to be introduced first. As shown in Fig. 9.5, the Q angle

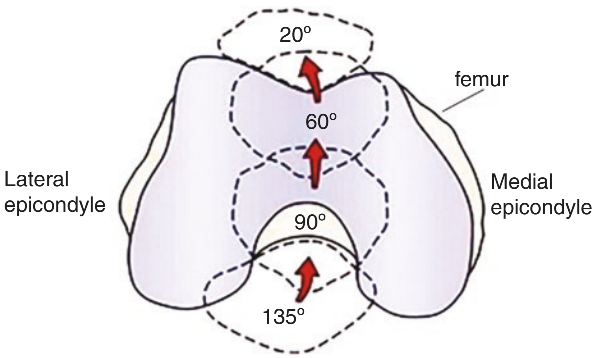
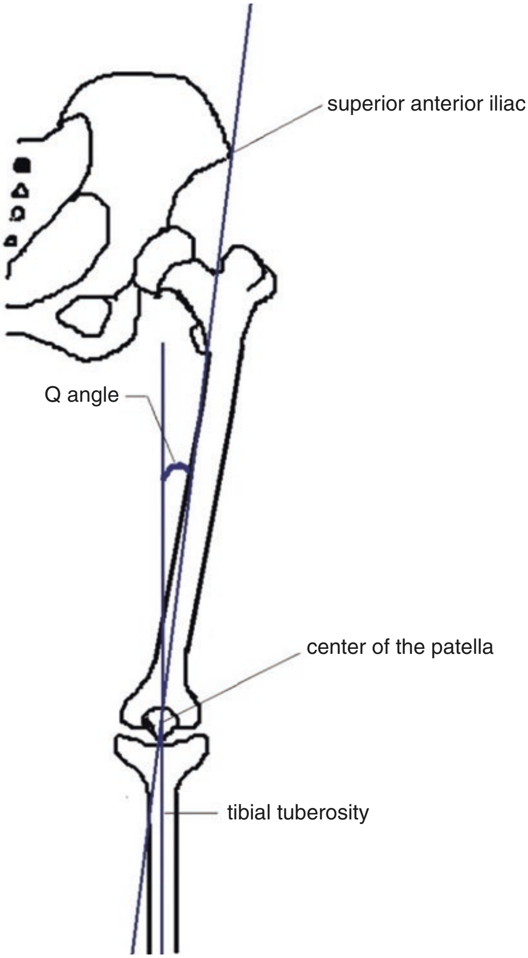


Fig. 9.4 Movement track of the patella during knee flexion

Fig. 9.5 Quadriceps angle



is the angle between a line that connects the center of the patella and the superior anterior iliac and another line that goes through the center of the patella and the tuberosity of the tibia. However, there is a significant difference in Q angle between males and females. For males, the Q angle at full knee extension ranges from 0° to 19°, while that for females is larger, ranging from 6° to 27° [39]. This is mainly due to the larger pelvis in females. Changes in the Q angle can help with understanding the mechanics of knee motions and patellar traction. During knee extension, the tibia gradually rotates externally and as a result the Q angle increases. Inversely, the Q angle decreases during knee flexion when the patella is pulled into the trochlea of the femur by a traction of horizontal forces from the oblique muscle. Thus, stability of the patella is largely dependent on the value of the Q angle and strength of the oblique muscle. Loss of stability may result in the patella tilting externally, resulting in wearing of the lateral patellar cartilage. The size of the Q angle can be affected by some factors, such as the size of the pelvic bone and connection between joint structures [40]. Abnormal joint structure such as excessive anterior tilt of the femoral head, excessive knee abduction, and other irregular rotations of the tibiofemoral joint could all result in an increased Q angle.

2 Pathomechanics of the Knee Joint and Clinical Treatment

2.1 ACL Injury and Clinical Treatments

2.1.1 ACL Injury

The ACL is frequently injured because of its critical role in constraining joint motion [41]. Most ACL injuries occurred in the absence of physical contact, and frequently during sport activities such as skiing, football, and basketball [41–43]. Females also have a higher incidence of injury than males [28]. Serious damage to the ACL can have a considerable impact on absence from work and the life quality. Long-term complications of ACL injury include secondary injuries to other knee structures such as meniscal tear, MCL injury that could induce degenerative changes to the cartilage both without and with surgical reconstruction [44, 45].

More than 75% of ACL tears are a complete rupture of the ligament, while the remaining 25% are partial tears [46]. Generally speaking, 1/3 of the patients had ACL reconstruction surgery, while a second 1/3 had delayed surgery. The remaining 1/3 would not require surgery. For the latter group, the ACL healed and the knee functions as well as those following surgery.

2.1.2 Conservative Treatment

Conservative treatments for ACL injury include plaster fixation, cryotherapy, muscle training, and strengthening of coordination [47]. With limited ACL function, anterior translation of the tibia may be restricted if there is adequate proprioception to contract the hamstring muscles [47]. However, although such conservative treatments are noninvasive and less stressful for the patient, the clinical outcome is relatively unpredictable and remains controversial. In a clinical report at 10–13 years after surgery, the ACL reconstruction group was reported to perform significantly better than the nonoperative group (conservative treatments) in maintaining involvement in sports [48]. In contrast, other studies have shown good patient satisfaction after 20 years following conservative treatments, though the objective measures indicated increasing knee degeneration [47]. When treated by conservative means alone, Noyes [49] reported that tears across one-quarter of the ACL body did not usually progress further, while one-half tears progressed in 50% of people and three-quarter tears progressed in 86% of people, eventually becoming a complete deficiency 7 years after treatment. Thus, the initial size of the ACL tear might be important for evaluating the expected clinical outcome of conservative treatment.

2.1.3 ACL Repair

The first attempts at repairing ruptured ACLs involved reconnecting the ligament remnants by sutures (termed a primary repair) to facilitate healing. However, in a 5-year clinical follow-up study on ACL primary repair, 53% of the patients had a reinjury, 94% suffered instability, and 71% had pain [50]. It was generally concluded that the poor results of primary repair could be attributed to the poor healing capacity of the ACL [51, 52].

Functional tissue engineering was introduced in the 1980s as a novel method to "grow tissues or organs from a single cell taken from an individual" [53]. With developments in functional tissue engineering, biological scaffolds such as extracellular matrix (ECM) have been used to reinforce ACL healing [54, 55]. ECM bioscaffolds serve to bridge the gap between the ACL remnants to facilitate the migration and proliferation of the cells and formation of the blood vessels, as well as the transportation of wastes, and thus consequently accelerate tissue formation and improve the healing of the ACL. Woo et al. [55] used ECM in sheet and gel forms to repair a transected goat ACL. The results showed continuous formation of neotissue in the ECM group while there was limited tissue growth in the control group (suture only). Also, the stiffness of the femur-ACL-tibia complex for the group using ECM was 2–3 times that of the group using suture repair alone.

Mechanical augmentation for ACL repair has also shown positive results for improving ACL healing. Fleming et al. introduced a form of suture augmentation where the sutures were fixed directly to the bone to reinforce the repaired ACL and reduce anterior laxity of the joint [56]. Fisher et al. [10] showed that, in comparison to traditional suture repair, mechanical augmentation of ruptured ACLs with sutures resulted in improved joint stability, higher load in the repaired ACL, and lower load in the medial meniscus, demonstrating the effectiveness of suture augmentation for restoring joint stability and protecting the medial meniscus. The subsequent introduction of biodegradable ACL grafts allowed for early reinforcement of the repaired ACL and gradual transferring of loading to the ACL as it heals, along with the concurrent dissolution of the graft [57]. Similarly, a degradable magnesium ring device was designed and used by Farraro et al. as mechanical reinforcement which was combined with biological reinforcement (ECM scaffold) for ACL repair to restore joint stability and to load the femoral insertion site to prevent disuse atrophy throughout the ligament healing [58].

2.1.4 ACL Reconstruction

Because of the bad capacity of healing of the ACL and the unsatisfying clinical outcomes of conservative treatments and repair surgeries, ACL reconstruction has remained the gold standard for treating ACL ruptures [59]. This involves replacement of the original ACL with a graft that is fixed into femoral and tibial tunnels to restore joint stability. There are a variety of sources for the graft, with the choice of using an allograft, autograft, or artificial graft. The bone-patellar tendon-bone complex (BPTB) and four-strand hamstring tendon (HS) have become the two most commonly used autografts, and the LARS artificial graft (LARS Company, France) is the most popular synthetic graft, with a market history of over 20 years [60, 61].

Satisfactory outcomes have been reported in the short term following ACL reconstruction. Previous study displayed no loss in range of motion of the knee 2 years after the surgery and limited postsurgical complications [62]. Another recent study [63] reported that 1 year following ACL reconstruction, the rate of graft failure was 6.5% for HS and 2.1% for BPTB, while the rate of returning to sports was 71% when using a HS graft and 78% when using a BPTB graft.

However, although ACL reconstruction shows positive results in the short term, long-term complications have been reported. First, the complex anatomy of the ACL insertions cannot be restored, and thus the rotational joint stability cannot be totally restored. Tashman et al. [64] measured knee kinematics during downhill running 4–12 months following ACL reconstruction and found that the anterior tibial translation for the reconstructed knee had no significant difference with the uninjured knee, but the reconstructed knee had larger external rotation (3.8° larger on average) and varus rotation (2.8° larger on average). Second, there could be donor site morbidity when using autografts. In a study by Salmon et al. [65], donor site related complications were reported to appear in 42% of the patients 13 years after ACL reconstruction, while 45% of patients suffered pain when kneeling. Third, there have been reports of enlargement of the bone tunnels and rupture of the grafts, which have an incidence of up to 72% and 22%, respectively [66, 67]. Bone tunnel enlargement can cause graft loosening, fixation failure, knee instability, and osteoarthritis, while a graft rupture directly calls for follow-up surgery. Other potential complications include loss of proprioception, graft wear, graft laxity [68], extension loss, and long-term osteoarthritis [69, 70]. A recent study reported an incidence of knee osteoarthritis of nearly 20% at 12 years following ACL reconstruction [45].

2.2 Meniscal Tear and Clinical Treatments

2.2.1 Meniscal Tear

A tear in the meniscus is a relatively common soft-tissue injury. A meniscal tear can cause pain, swelling, and joint instability, and long term could induce knee osteoarthritis due to increased stresses on the hyaline cartilage. Englund et al. [71] assessed the integrity of the menisci in 991 subjects using a 1.5 T MRI and found that 19% of females and 56% of males had some level of tearing, and over 60% of subjects with osteoarthritis had a meniscal tear. Among the subjects without any symptoms of knee pain, aching, or stiffness, 60% were found to have a meniscal tear.

There are a variety of types of meniscal tear which may be characterized based on the location and appearance: horizontal tear, flap tear, radial tear, root tear, vertical tear, oblique tear, and complex tear.

Horizontal tears are defined as occurring parallel to the tibial plateau in the region that divides the meniscus into superior and inferior halves. A flap tear occurs where there are displacements of torn fragments toward the articular space. A radial tear is vertical to the tibial plateau and transects the longitudinal collagen bundle. This type of meniscal tear impairs the axial strength of meniscus and might increase the risk of meniscal extrusion [72]. A meniscal root tear is defined when a radial tear happens within 1 cm of the bony tibial attachment. A vertical tear runs parallel to the longitudinal collagen bundle and divides the meniscus into peripheral and central portions. Oblique tears are oriented oblique to the longitudinal collagen fibers. Complex tears are defined as a combination of two or more types of tears occurring simultaneously. According to Englund et al. [71], among 308 subjects with meniscal tear, 40% were horizontal tears, 37% were complex tears, 15% were radial tears, 12% were oblique tears, 7% were longitudinal (vertical) tears, and 1% were root tears.

2.2.2 Meniscal Repair

Where appropriate, repairing the meniscus with sutures is often a good option for restoring normal mechanical function while maintaining proprioception and vascularity. In a 2-year follow-up study on 280 meniscal repairs [73], successful results were obtained in 252 knees, with a total failure rate of 10%. Of those patients that were deemed successful, there was an absence of any tenderness of the joint line, blocking or swelling. However, whether to perform a meniscal repair could be largely dependent on the tear type and the surgeon's individual ability and experience. Vascularity is only provided at the periphery of the menisci (up to 25% of the width, termed the red-red zone) (Fig. 9.6) as well as at its horn attachments [74], and thus the capacity of healing of the meniscus is rather poor in the middle portion (termed the white-white zone) (Fig. 9.6). Previous studies demonstrated how tears on the posterior root of the medial meniscus could be repaired to restore the normal joint loading profile [75]. If a vertical tear is located peripherally, meniscal repair using sutures is usually the preferred treatment [76]. Other types of meniscal tears, particularly radial tears, are more difficult to repair and heal. For these difficult

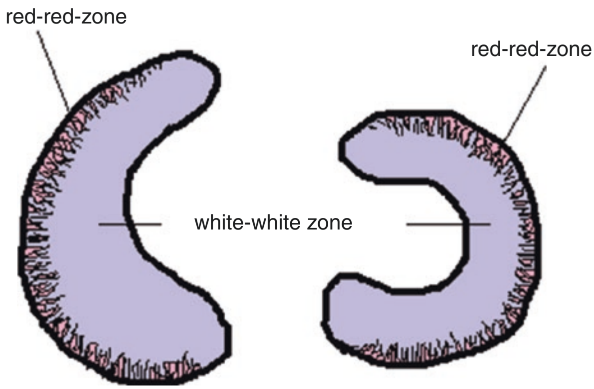


Fig. 9.6 Illustration of the red-red and white-white zones of the menisci

tears, meniscectomy or meniscal replacement may be a more suitable option. It was reported that around 25% of meniscal tears could be repaired in patients younger than 40 years old [77].

2.2.3 Meniscectomy

Meniscectomy is a surgical procedure to partially or completely remove a damaged meniscus. A clinical follow-up study by Roos showed that undergoing meniscectomy increased the risk of tibiofemoral osteoarthritis by 14 times at 21 years after the surgery [78]. Ihn et al. performed cadaveric testing using pressure films [79] and found that following meniscectomy, the peak contact stress in the knee joint increased because of the reduced contact area.

Compared with partial meniscectomy, total meniscectomy could result in a smaller articular contact area, which may speed the development of osteoarthritis. Meniscectomy could also have a considerable effect on the ligaments, with one study [65] reporting a sixfold increase in graft rupture 13 years after undergoing meniscectomy and ACL reconstruction. Conversely, forces in the meniscus could be doubled after transecting the ACL [80].

In comparison to medial meniscectomy, lateral meniscectomy is reported to have more postoperative complications [81]. As mentioned previously, the lateral meniscus shares more load for the knee compartments. Thus, removal of the lateral meniscus could be expected to have a more pronounced effect on stress distribution in the joint. Also, the lateral tibial plateau is convex in contrary to the concave medial plateau; therefore, removal of the lateral meniscus would cause more point contacts and increase the peak articular contact stress. A clinical study [82] on meniscectomy reported good results in 80% of the patients following medial meniscectomy while only 47% of patients who underwent lateral meniscectomy reported similar results.

2.2.4 Meniscal Replacement

If a patient develops complications such as joint pain from early degeneration after meniscectomy, they may be considered as a candidate for meniscal replacement.

Allograft

Meniscal allografts are obtained from donors and should contain no living cells [77]. If donor cellular material remains in the graft, a dense cartilage matrix would be used to lock the materials so that the allograft does not elicit any immune response.

Meniscal replacement was first introduced in the 1980s [83]. A previous study [84] reported a total failure rate of 29% for allograft transplants 13 years after the replacement surgery, while the overall Lysholm score improved from 36 to 61 comparing with the preoperative situation, showing a promising potential for relieving pain and postponing the need for knee arthroplasty in younger patients [85]. Similar results have been reported by other researchers [86], whereby meniscal replacement has been shown to prevent further narrowing of the joint space, reduce knee pain, and halt further degeneration of the cartilage.

Meniscal Bioscaffold

The red-red zone of the meniscus has a vascular supply while the middle section (white-white zone) does not. In a partial meniscectomy, if the red-red zone is well preserved, an artificial scaffold can be sutured to replace the missing tissue and restore the mechanical function of the meniscus.

The Menaflex Collagen Meniscal Scaffold (Regen Biologics Inc., New Jersey, USA) and the Actifit Scaffold (Orteq, London, UK) are two of the most commonly used meniscal scaffolds on the market. The former is made of Achilles tendon of bovine that is pressure heat molded into the shape of a meniscus. A clinical follow-up study demonstrated significantly increased tissue formation with the use of this scaffold over a partial meniscectomy [87], and more activity was regained postoperatively in the scaffold group than the meniscectomy group. The Actifit scaffold is a porous biodegradable scaffold made of polyurethane that allows neo-tissue to gradually replace the scaffold material over time. A 1-year biopsy study showed viable “meniscus-like” tissue replacing the scaffold material and a considerable reduction in patient pain [77].

2.3 Knee Osteoarthritis and Treatment Options

2.3.1 Knee Osteoarthritis

Knee osteoarthritis is a common disease in middle-aged and older people, presenting as progressive degeneration of the joint cartilage, including osteophytes formation, cartilage loss, sclerosis of subchondral bone, degeneration of menisci, laxity of ligaments and alterations in the synovium and fat pad [23]. There are numerous factors that may lead a person to develop osteoarthritis, to name a few, cumulative high weight-bearing, poor posture during gait, muscle weakness, and sudden temperature changes [88–90]. With knee osteoarthritis, people suffer from stiffness, pain, and loss of knee function.

For patients in the early stages of arthritis, basic and conservative treatments are often used, such as nonsteroidal anti-inflammatory medication, localized corticosteroid injections, and weight loss [91]. As the osteoarthritis progresses, destruction of the matrix occurs and the external loads exceed the physiological limit of the cartilage, resulting in asymmetric wearing of the cartilage and subsequent bone loss. Generally, once the pain reaches a certain level, the patient and physician may opt for surgical knee osteotomy. This may be a high tibial osteotomy or fibular osteotomy. However, under severe loss of cartilage and/or bone and consequent intolerable pain and dysfunction of the knee joint, the only viable option is to undergo knee arthroplasty, during which the joint surfaces are replaced by prosthetic components. The ultimate goal of knee arthroplasty is to replace the problematic knee joint with a prosthetic knee in order to restore the joint functionality and relieve the pain.

2.3.2 High Tibial Osteotomy (HTO)

Although osteoarthritis can present in any of the knee compartments, a large portion of the cases reported occur in the medial compartment of the tibiofemoral joint [92], with fewer cases being reported for the lateral side or the patellofemoral joint [93]. When osteoarthritis appears in the medial side, the medial joint space is usually narrowed because of the loss of cartilage and bone, resulting in abnormal force transmission along the lower limb.

For patients who are younger and more active, high tibial osteotomy (HTO) is often chosen for early and moderate osteoarthritis. In this case, part of the tibial bone is removed to relocate the joint surface and to restore joint alignment. For medial arthritis, more bone would be removed at the medial side, changing the joint alignment from varus to be slightly valgus, thus decreasing medial joint pressure and slowing down the degeneration of the medial cartilage. This may postpone or avoid the necessity for knee arthroplasty. Previous studies comparing clinical results of HTO and unicompartmental knee arthroplasty (UKA) concluded that UKA presents lower revision rates, less postoperative complications, and less postoperative pain. However, HTO allowed for a better range of motion, which could be more satisfactory for highly active patients [94].

2.3.3 Fibular Osteotomy

The fibula bone is generally considered to be less critical for joint functionality than the other long bones of the lower limb. The fibula provides an origin for several muscles of the knee joint and carries axial weight-bearing loads ranging from 6.4% [95] to 16.7% [96]. It has been freely used as a source for grafts and as a vascularized transplant to bridge large bony defects for conditions such as congenital pseudarthrosis of the tibia [97], tumor resection [98], nonunion [99] and grafting operations in case of femoral head necrosis [100].

Proximal fibulectomy (fibular osteotomy) has been demonstrated by several studies to improve the restoration of joint functionality and reduce pain in patients with OA. It is a simple but effective procedure, with the number of patients with a fibulectomy in China surpassing 1000 in 2015 [101]. To perform a fibular osteotomy, the proximal part of the fibula is first exposed, followed by an osteotomy with a length of 2–4 cm at 7–8 cm from the proximal fibula head. In a follow-up study on 110 patients over a 2-year period, Yang et al. [102]

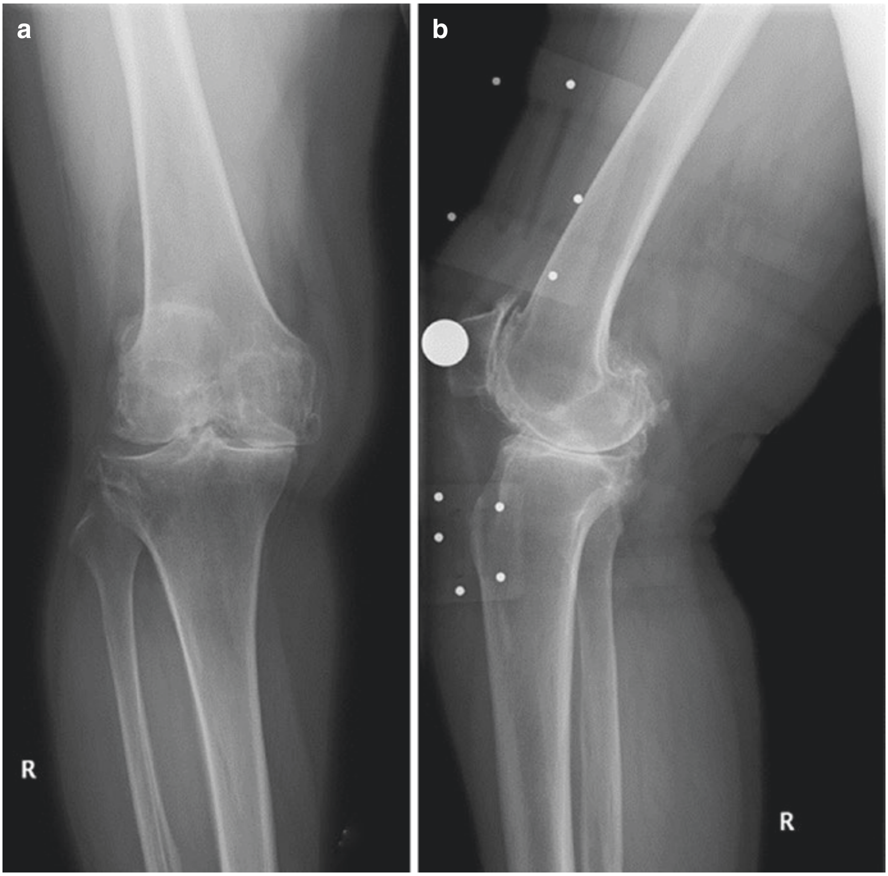


Fig. 9.7 Preoperative radiographs; anteroposterior (**a**) and lateral (**b**)

reported significant improvement in alignment (Figs. 9.7 and 9.8) and function of the knee, as well as relief of pain following the fibular osteotomy. Zou et al. [103] investigated 92 patients with varus knee OA treated by either fibular osteotomy or HTO and found that the outcomes of fibular osteotomy were superior to HTO in either short term or long term. Nie et al. [104] reported on 16 patients with knee OA in the medial compartment who underwent proximal fibulectomy and found that there were significant improvement in VAS pain and HSS score 1 day and 6 months after surgery. In addition, the hip-knee-angle (HKA) improved and remained stable for 3 months and there was an inverse relationship between the overall peak knee adduction moment (KAM) and HKA. Yazdi et al. [105] evaluated the effect of partial fibulectomy on the articular contact pressure in cadaver knees and it was demonstrated that partial fibulectomy decreased the medial articular pressure and increased that in the lateral compartment. Although the procedure is relatively simple and there are good clinical results, it may

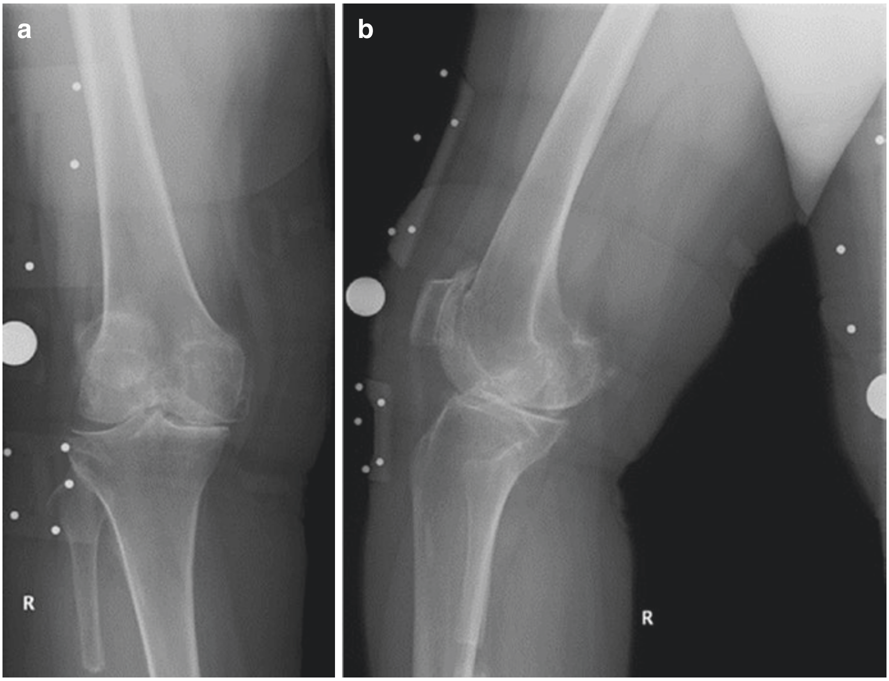


Fig. 9.8 Immediately postoperative radiographs; anteroposterior (**a**) and lateral (**b**)

cause adjacent peroneal nerve injury. Sandoval et al. [106] followed up 116 patients with proximal fibulectomy for 2 years and reported that only 9.4% suffered complications, mainly for neuropraxia of the peroneal nerve, hematoma of the wound and infection. While fibulectomy has generally been shown to be effective, it is not clear exactly how the procedure relieves knee pain. Zhang et al. [107] stated that nonuniform settlement and bilateral degeneration of the plateau leads to varus knee, while with an osteotomy of the fibula can modify this situation. Weakened support from the lateral fibula results in a correction of the varus deformity, and consequently shifted the loading from the medial compartment to the lateral, which eventually leads to relieved pain and restored joint functionality.

Qi et al. [108] hypothesized that a reduction in lateral muscle contraction after fibulectomy may cause a rebalance in the resultant joint moment, making the contact forces in the medial compartment shift laterally and decrease in magnitude.

According to Qi [108], the balance of joint forces in the coronal plane of a varus knee may be gauged by the following equations (Fig. 9.9):

$F_{LM} + F_B + F_{MM} = F_{JL} + F_{JM}$, where F_{JM} is the medial joint contact force, F_{JL} is the lateral joint contact force, F_{LM} is the lateral resultant muscle force, F_B is body weight due to the gravity, F_{MM} is the medial resultant muscle force, and F_{JM} can be ignored due to varus knee.

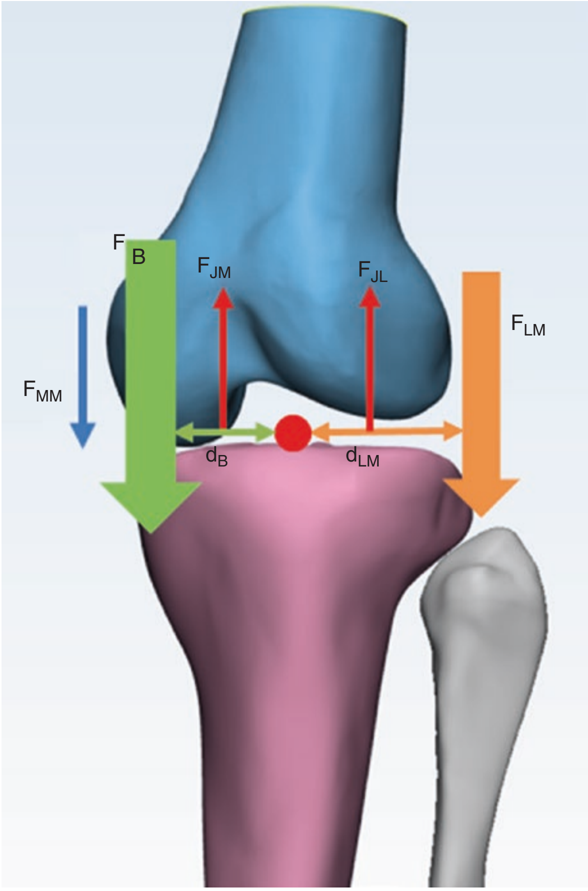


Fig. 9.9 Before operation

$d_{LM} \times F_{LM} = d_B \times F_B + d_{MM} \times F_{MM}$, where d_{LM} is the moment arm of lateral resultant muscle force to knee joint center, d_B is the moment arm of body weight to the knee joint center, and d_{MM} is the moment arm of medial resultant muscle force to the knee joint.

After removing a portion of the fibula (Fig. 9.10), the lateral muscle (F'_{LM}) is released which decreases the valgus moment ($d_{LM} \times F'_{LM} < d_B \times F_B$). To compensate, the varus resultant moment is reduced by shifting the body weight laterally which subsequently decreases the moment arm of body weight ($d'_B < d_B$). Meanwhile due to the decrease in the lateral muscle force (F'_{LM}), the overall knee joint forces also decrease ($F'_{LM} + F_B = F'_{JL} + F'_{JM}$).

In summary, proximal fibulectomy is an innovative procedure to treat medial knee OA without the need for resecting the knee joint. Studies performed to date lack sufficient quantitative biomechanical evidence to recommend using this procedure as standard practice. Further clinical follow-up studies and comparative analyses to other common treatment methods are required.

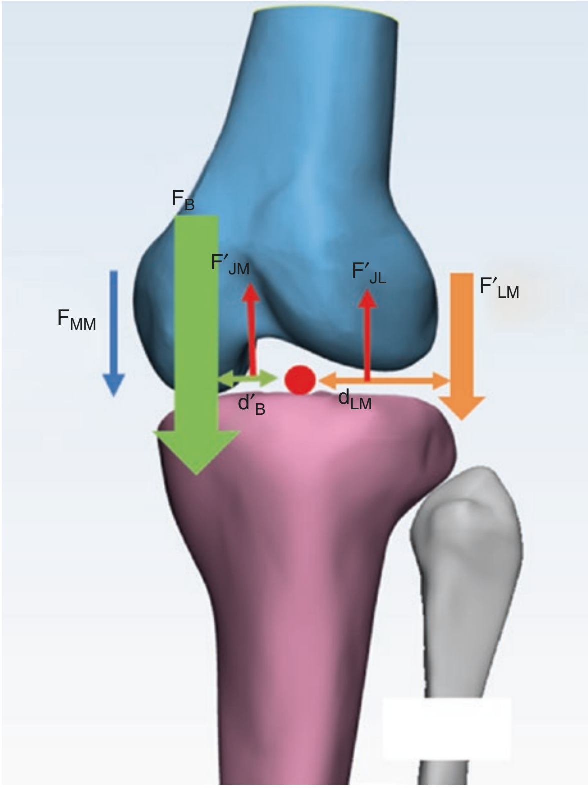


Fig. 9.10 After fibular osteotomy

2.3.4 Knee Arthroplasty

Knee arthroplasty is often considered when conservative treatments and less invasive bone osteotomies are either unsuccessful or are not considered a viable option because the natural cartilage has reached an irreparable state. Knee arthroplasty can offer pain relief and help regain normal joint function. The knee prosthesis generally consists of the femoral, tibial, and patellar components. The tibial component usually contains a polyethylene liner inserted into a metal baseplate.

Knee arthroplasty may be performed as a unicompartmental knee arthroplasty (UKA) or a total knee arthroplasty (TKA). For TKA, there are generally two types of knee prosthesis in clinical use: cruciate-retaining type (CR type) and posterior-stabilized type (PS type).

Unicompartmental Knee Arthroplasty (UKA)

UKA is an operation used to relieve arthritis by replacing the articular surface of either the medial or lateral condyle without sacrificing the cruciate ligaments and the patella, as opposed to total knee arthroplasty (TKA) which replaces both knee compartments at the same time. In theory, because UKA retains more soft tissues, it can better restore knee functionality and stability than TKA and would allow the patient to retain a better ontological feel. In the early years following the introduction of unicompartmental procedures, the failure rate was quite high due to poor surgical techniques and poorly designed components. The average revision rate of UKA at 2 and 6 years has been reported at 20% and 28%, respectively, in 1970s [109, 110]. With successive improvements in the design of components, mature clinical techniques using minimally invasive surgery have significantly improved the clinical success rate with appropriate patient selection [111]. The survival rate of UKA at 2 and 5 year is 98.7% and 95.5%, respectively, and the knee function score remains satisfactory [112]. However, the long-term survival rate is significantly lower than that of TKA [113]. Failure rates of UKA vary across literature and for different regions [114–117]; 27% (10 years) in Finland, 12.1% (8 years) in UK, and 4.6% (10 years) in Japan. The success of UKA is dependent on many factors, including the age and activity of the patient, surgeon's experience, and the type of implant used.

UKA has been used in clinic for over 60 years. In the early 1950s, McKeever [118] inserted metal plates into a patient to replace the damaged articular cartilage surface, becoming one of the first documented cases of UKA. By the 1960s the procedure had reached mainstream practice thanks to improved designs and the use of alternative materials. Gunston [119] developed the first unilateral metal-on-PE bearing, which is still the most common material combination in use today. The 1970s saw the introduction of non-constrained prostheses that increased the activity of the joint and could achieve a greater joint range of motion than TKA prostheses [120]. In the 1980s, Goodfellow introduced the idea of an Oxford active platform into the design of single condyle prostheses (Oxford unicompartmental knee replacement, OUKR) which increased the articular contact area and reduced the contact stress [121]. In recent years, custom-made UKA prostheses (for example, iUni by ConforMIS Inc.) developed from CT scans and manufactured by 3D reconstruction and printing have been introduced. As the shape of the prosthesis is designed according to the individual anatomy of the femoral condyles, custom prostheses can provide better alignment and more natural movement. However, given that custom implants are a recent development, the long-term outcome is relatively unknown. According to the American Joint Replacement Registry [122], UKA accounted for about 3.2% of primary knee replacement procedures from 2012 to 2017, and there was a downward trend year on year. This same trend has also been observed in Australia [123] and Sweden [124], with the former showing a decrease in UKA from 14.5% in 2003 to 5.1% in 2016. However, since 2017 the trend has reversed and there has been a slight uptake in UKA procedures in Australia [123], Sweden [124], and Britain [125].

There are a number of potential causes of failure of unilateral knee prostheses, including degeneration of the adjacent tibiofemoral joint and patellofemoral joint, implant loosening, mechanical injury, malposition, and infection. Among them, aseptic loosening and joint disease (including contralateral arthritis) are the most common causes of failure [123–125]. Typical factors leading to loosening of the prosthesis are malalignment of the components, undercorrection of a preexisting deformity, deficiency of the ACL, improper tibial slope, and dislocation of the bearing in mobile designs. Each of these factors may cause asymmetrical loading on the prosthesis and therefore produce more wear particles, leading to further osteolysis. Joint diseases after UKA, such as the progressing of OA in the other knee compartments, can be caused by overcorrection of the mechanical axis or impingement due to an oversized femoral component [126].

At present, biomechanical studies of unicondylar knee prostheses typically focus on the effect of the mechanical axis of the knee joint on the medial or lateral articular stress. If the stress of the articular surface is not evenly distributed, it may accelerate implant failure [127]. In general, the stability of ligaments and the amount of angular deformation must be carefully assessed preoperatively. Medial arthritis often first appears in the anteromedial region of the knee joint, and then gradually extends to the femoral condyles and intercondylar fossa. Inflammation of the knee caused by cartilage degeneration and the presence of intercondylar fossa osteophytes alters the biochemical and mechanical environment of the ACL. Degeneration of the ACL subsequently changes the kinematics of the knee joint, leading to further degeneration of the posteromedial region. In the early stages of medial knee arthritis, if the ACL does not display functional deficiencies, UKA can be chosen to slow down regression of the ACL and the contralateral compartment. However, if the functionality of the ACL is impaired, for example, through rupture or resection, then the stability of the knee joint may be affected. In such situations UKA is not a viable option.

Total Knee Arthroplasty (TKA)

In the operation of TKA, all of the articular surfaces in the knee are replaced with prosthetic components. The two most common types of total knee prostheses in clinical use today are cruciate-retaining (CR) and posterior-stabilized (PS). Both types require the removing of the ACL, but with CR prostheses the posterior cruciate ligament (PCL) is retained, while PS prostheses also require a resection of the PCL. With the PS knee, a femoral cam and a tibial post interact as a functional replacement for the PCL. When the femoral component slides anteriorly, the cam contacts the tibial post, forcing the femoral component to roll back along the tibia. This mechanism is designed to prevent excessive anterior femoral translation and help to maintain joint stability.

The anterior region of the tibial liner in TKA is often designed to be higher than the posterior part in order to avoid excessive anterior femoral translation and increase the contact area of the tibiofemoral joint. This stabilizes the knee and decreases contact stress on the components [128].

The polyethylene liner in the tibial component may be anatomically shaped ("anatomical design") (Fig. 9.11b) or be a simpler nonanatomical design (Fig. 9.11a). For the nonanatomical design, the medial and lateral part of the polyethylene liner are often symmetrical, concave, and have the same curvature radius [128]. In this case, the curvature of the femoral component may be also designed to conform to

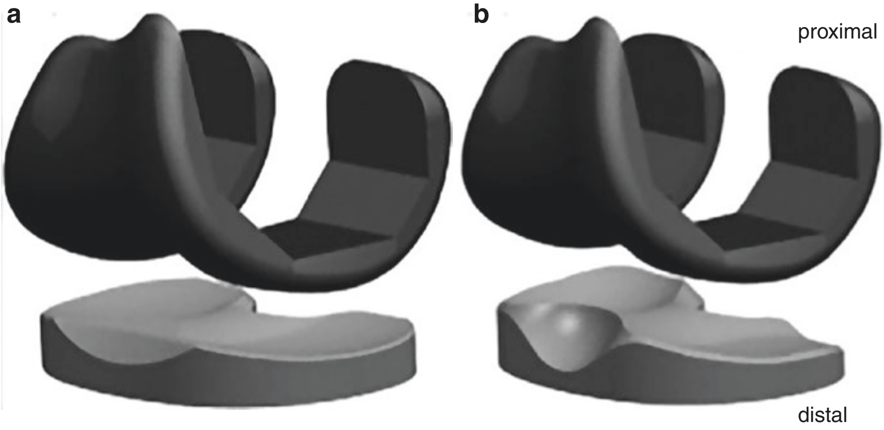


Fig. 9.11 Different designs of the polyethylene liner (**a**) nonanatomical design and (**b**) anatomical design [129] (Reprinted with permission from Elsevier, license number: 4687410127428)

the tibial liner in order to maintain the knee stability and prevent component dislocation [128]. However, the normal knee shape is asymmetrical and this nonanatomical design could induce abnormal knee kinematics such as inadequate rollback of the femur and insufficient internal rotation of the tibia. Thus, anatomical prosthesis designs aim to mimic the morphology of the intact knee to restore the kinematics of the knee. Liu et al. found that [129] changing the posterolateral region of the tibial liner to be convex could better restore the posterior translation of the lateral femoral condyle and the internal rotation of the tibia through joint flexion from 0° to 140°. A study by Moonot et al. [130] demonstrated that asymmetrical tibial liners with a highly conforming medial aspect and a moderately conforming lateral aspect could adequately restore the internal tibial rotation and reduce excessive anterior movement of the lateral femoral condyle, thus reducing wear on the liner and increasing implant longevity.

When considering the type of TKA implant to be used, it is still controversial whether the PCL needs to be retained for proper knee functionality [131]. CR knees require an intact and functioning PCL, and so this must be carefully evaluated beforehand. Otherwise, if the PCL is compromised then joint stability could be at risk. In this case a PS knee is more desirable. However, when using a PS design, joint stability is greatly affected by the design of the tibial post and the femoral cam, and this would also have a considerable effect on joint kinematics and kinetics.

If the cartilage on the patella is degraded and may compromise knee functionality, then there is the option of replacing the cartilage with a smooth polymer surface (patellar replacement). However, Cadambi et al. reported that replacing the patellar surface reduces the success rate of TKA [132]. Huang et al. also showed that tensile stress on the patellar bone increased after the surface was replaced [133]. Rand [134] stated that introduction of the patellar component could alleviate anterior knee pain of the patients. However, the revision rate of TKA is still relatively high because of the patellar component related complications, such as infection, fracture, subluxation, wear, and loosening of the patellar component [135, 136]. Obese patients are also not recommended to undergo patellar arthroplasty because the higher loading could increase the risk of loosening and fracture [137].

3 Summary

In this chapter, the basic anatomy and functionality of bony structures, menisci, cartilage, tendons, and muscles in a normal knee joint were presented, accompanied by a discussion of the biomechanics of the patellofemoral and tibiofemoral joints. Pathological changes to the structures of the knee could alter the joint kinematics, which in long term could induce secondary injuries to the other surrounding tissues. Clinical treatments should be carefully chosen to best rebuild the normal biomechanics and functionality of the knee.

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