The Biomechanics of the Knee Joint

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Peter Theobald, Samih Tarabichi, and Mohamed Elfekky

The field of biomechanics serves to link the seemingly disparate specialisms of biology and mechanical engineering. Application of biomechanics to the musculoskeletal system is documented from the early twentieth century, as innovators and pioneers began investigating synovial joint behavior with a view to understanding the link between its structure and function. These early attempts, to quantify mechanical parameters of the mammalian synovial joint, provided the building blocks for the current researchers to explore equivalent human parameters, in vivo. Biomechanical engineering now extends to describe a breadth of mechanical engineering applied to the human body, extending from the micro-scale cellular biomechanics, to the gross-scale sports biomechanics. The field is now overseen by international bodies including the American Society of Biomechanical Engineering, the European Society of Biomechanics, and the International Society of Biomechanics in Sport.

Comprising the patellofemoral, tibiofemoral, and, to a lesser extent, the tibiofibular joint, highly complex biomechanics underpin knee joint behavior and its ability to withstand intense mechanical loading for typically 60–70 years. Musculoskeletal biomechanics allows quantification of these multifactorial interactions, enabling an enhanced understanding of the relationship between the tissue structure and its function, for example, quantifying the friction when the fibrous/fluidic articular cartilage surfaces compress against and then translate past, one another. First, however, appreciating the fundamental biomechanics will assist in understanding these more complex behaviors.

23.1 Musculoskeletal Mechanics

Musculoskeletal soft tissues are typically collagenous-based structures, bathed in a fluid of water and salts [1]. The interaction of these solid and fluid components produces a complex behavior when the tissues are compressed, extended, or subject to a shear loading [1]. This is caused by the fluid resisting the deformation of the loaded, fibrous network. Fluidic resistance increases when the rate of deformation is increased, meaning the tissue behavior depends on the loading speed. This is called viscoelastic behavior and, in soft tissues, allows greater deformation at lower strain rates. Key traits of any viscoelastic material are its creep and stress relaxation behavior. The former describes the extension of the soft tissue when exposed to a constant load, while the latter

P. Theobald
School of Engineering, Cardiff University, The Parade, Cardiff, UK

S. Tarabichi
Orthopaedic Surgery, Dubai, UAE

M. Elfekky (✉)
Hatta Hospital, Dubai Health Authority, Dubai, UAE

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describes the reduction in load when maintaining a constant length. The precise stress relaxation and creep behavior of a tissue are governed by its structural and fluidic compositions [2, 3].

Articular cartilage is perhaps the single most important structure in maintaining synovial joint longevity, given that it is the ease of the femoral surface translating against the tibial surface that is common with the healthy joint. Wet articular cartilage comprises 20% collagen and 5% proteoglycans, which form strong structural networks that resist mechanical loading [3]. These networks, however, are unevenly distributed through the cartilage thickness, with the collagen fibers in the upper 10% aligned parallel to the cartilage surface. Collagen fiber bundles in the deepest 30% are radially orientated, while the middle 60% has the lowest fiber density. This tissue network is then bathed in water and salts, which comprises the remaining wet weight. It is the flow and frictional drag of this fluid that govern the articular cartilage's viscoelastic behavior [2].

The tendons and ligaments of the synovial joint serve to transmit motion and provide stability, respectively. In the knee, the tendons overlay the anterior and posterior aspects of the joint, transferring contractile force from the muscle groups of the upper leg and so generating motion. Ligaments constrain tibiofemoral translation and rotation, so providing the requisite skeletal rigidity (Fig. 23.1). Wet tendons and ligaments comprise approximately 30% collagen and 70% water. Collagen forms approximately 80% dry weight, with the vast majority comprising collagen type I [4]. Collagen fibrils are inherently wavy, or "crimped," providing the unloaded tendon with an innate flexibility. Several fibrils are grouped, embedded within an amorphous matrix and then wrapped in connective tissue, to form a collagen fiber. An accumulation of fibers forms a bundle, with the matrix enabling force transfer force along its length. Multiple bundles are then wrapped to produce a fascicle and multiple fascicles gathered to form the final tissue, which itself is wrapped in a loose, connective tissue to facilitate gliding against neighboring structures. While a tendon or ligament can be broadly considered analogous to a multi-strand rope, being stiff in tension though offering negligible resistance in compression, their behaviors are subtly different as the collagen fibers in the former tend to be better aligned, as they carry greater uniaxial

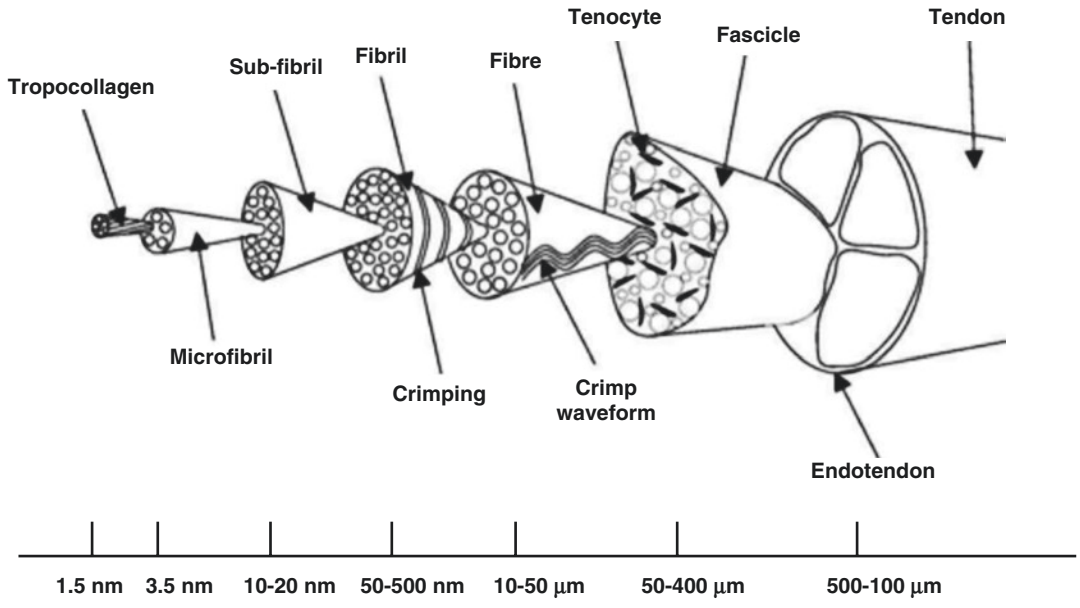

Fig. 23.1 Demonstrating how the collagen fibers are bundled together to form a tendon [4]

load [5]. Ligaments experience greater off-axis loads, meaning its fibers are less aligned.

23.2 Knee Kinematics

Knee functionality focusses on achieving and maintaining joint mobility and joint stability. Synovial joint movement is described by quantifying the extent of translation and rotation, or the range of motion, in the three orthogonal planes. Tibiofemoral rotation is the main source of knee joint mobility, achieving approximately 160° flexion extension. During the initial 30° flexion, internal tibial rotation also occurs, guided by the medial and lateral collateral ligaments. This represents the reversal of the "screw-home mechanism," with external rotation during the last 30° extension providing greater stability when the knee is fully extended [6]. This is achieved as the medial condyle is larger, generating rotation as it translates 5–10 mm across the tibial plateau. Lateral condylar translation is less pronounced. Full flexion achieves a posteriorly located tibio femoral contact region. In the frontal plane, 6–8° varus-valgus is evident in extension and is accompanied by 1–2 mm medio-lateral translation.

The need to constrain motion is critical to achieving an efficient gait. Excessive motion in the lower limb joints may cause greater medio-lateral or antero-posterior displacement in the overall center of mass, necessitating the generation of greater muscular force to retain balance. Knee motion is constrained by the two cruciate ligaments forming a "modified four-bar linkage mechanism," where the other two bars are represented by the bone that links the femoral insertion sites and the tibial insertion sites. During joint motion, the angles between the bars change, though stability is maintained as the four bars remain a broadly constant length [7]. Conventional linkages have four stiff bars. Marginal ligament extension enables the internal-external rotation, associated with the screw-home mechanism.

By balancing the conflicting demands of mobility and stability, the biomechanical properties of ligaments and tendons underpin the functionality of the natural joint. This behavior is typically described via a load–extension graph (Fig. 23.2). To plot these data independent of cross-sectional area and thereby enable comparison across tissues and individuals, the stress (the load divided by the area) and strain (the extension achieved, relative to the original length) are plotted to establish a correlation. Plots typically exhibit an initial toe region, where the crimped collagen fibrils achieve relatively great strain for a given stress. Additional extension then requires a greater load, due to the taut fibers providing greater resistance. During this phase, the stress–

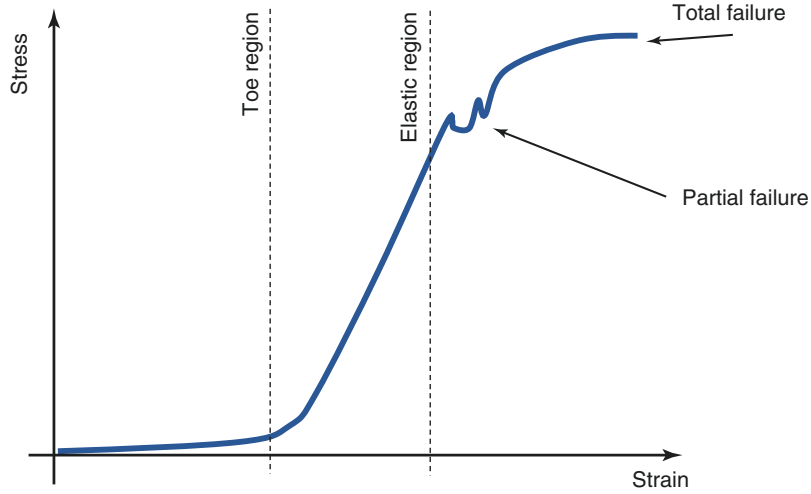

Fig. 23.2 The stress–strain relationship that is commonly observed when tensioning a tendon (author's drawing)

strain correlation will remain consistent, until such time as individual fibrils begin to fail. This failing will be represented by a reducing stress, caused by a relative reduction in the load required for extension. The stress will then continue to rise as greater load is required for ultimate failure, though distributed across fewer fibrils, until the tissue finally ruptures [8].

23.3 Knee Stability

The physical behavior and clinical observations of the healthy and pathological knee are often described by simplified terminology and subjective measurements, yet motion at the knee joint is a complex mechanical phenomenon. Terms such as instability and laxity are used interchangeably to describe a pathologic deviation from normal knee behavior; however, the definitions of these terms are ambiguous and do not provide objective information regarding knee kinematics and function.

Stability is provided by a combination of static and dynamic structures that work in concert to prevent excessive movement, or instability, which is inherent in various knee injuries. The anterior cruciate ligament is a main stabilizer of the knee, providing both translational and rotatory constraints. From a biomechanical standpoint, laxity describes the passive response of a joint to an externally applied force, whereas instability is typically expressed by the patient and constitutes a functional measure. Laxity and instability can be better quantified by understanding the stiffness of a joint complex [9]. Tissues with greater stiffness have the effect of limiting overall joint motion, although abnormally high tissue stiffness can over-constrain joint motion, as is possible in procedures such as extra-articular tenodesis. An increased laxity is not tantamount to knee instability nor an inferior outcome, as patients compensate differently and experience differing levels of disability.

23.4 Tibiofemoral Joint Kinematics and Forces

Static analysis can be used to estimate the compressive forces acting on the tibiofemoral and patellofemoral joints. This principle freezes a dynamic event, enabling analysis of a single time-point by considering only the predominant forces. Static analysis is typically used to investigate relative joint reaction forces (JRF), in this instance typically the tibiofemoral force, by calculating the resultant vector of the ground reaction force (GRF) and muscular force. This "inverse" method enables comparison of different every-day activities, or to track rehabilitation after joint replacement surgery. The GRF is measured using force plates, which may be embedded in the floor in a laboratory environment or used as a portable device for in situ testing. GRF represents the equal and opposite reaction of the ground when loaded by the human body. The muscular force can then be calculated by resolving the moments about the center of joint rotation (i.e., where the JRF acts through), enabling the JRF to be calculated by summing all forces to zero. When the knee is in full extension, there is no resultant moment nor muscular contraction, meaning that in double-leg stance, each tibiofemoral joint is exposed to one-half total body weight (the lower leg weight, approximately 5% BW, is ignored for simplicity). A static analysis of more complex scenarios, for example, walking, demonstrate that the JRF is greatest at the greatest knee flexion angle, which is approximately 20° when walking on the flat. Different peak flexion angles when ascending (~60° flexion) and descending stairs (~85° flexion) are examples of producing different JRFs during tasks that may seem similar.

23.5 Patellofemoral Joint Kinematics and Forces

Patellofemoral range of motion is predominantly defined by the patella's posterior anatomy and that of the underlying femoral condyles. The

patella articulates with the lateral femoral ridge as it approaches full extension, ultimately fitting between the medial and lateral condyles. It glides into the intercondylar notch at full flexion. The patellofemoral JRF increases during flexion due to the increasing patella tendon force, approaching 4 × BW during stair climbing. The increased patella tendon force also increases the patellofemoral contact area, which serves to better dissipate the greater JRF and so moderate joint stress [10].

23.6 Joint Tribology

While mobility and stability are the key traits of the knee and wider synovial joint family, maintaining these functionalities over a lifetime is highly demanding. The longevity of the synovial joints is remarkable given the amount of translation and rotation, which risks degrading the surfaces as they move past one another. Further still, the musculoskeletal system is frequently exposed to loads multiple-times body weight, acting on very small contact areas between the articulating surfaces in the lower limbs. This can generate extreme contact pressures. The very slow cartilage repair process means that the synovial joint needs to be hard-wearing and resilient, as structural damage is likely to encourage further degradation.

The ability of the healthy articular cartilage to withstand this demanding mechanical environment has long been investigated, with studies from the early twentieth century, reporting the oozing of synovial fluid from the tissue surface when exposed to a compressive load [11]. It was, and is still, believed that this process is of critical importance to protecting the interacting surfaces as the femoral condyles glide over the tibia, minimizing articular cartilage wear. The challenge of explaining how this process works, however, remains unsolved, limiting the ability of biomechanical engineers to design replacement components. Biomechanical engineers have long been seeking solutions from their mechanical-based colleagues, who can mathematically predict the extent of lubrication between synthetic bearing surfaces, enabling design changes that leverage enhanced performance.

The synovial joint has a small volume of synovial fluid, produced via the synovium and retained within the joint by the synovial capsule. When there is no motion between the articulating surfaces, it is typical for the two articulating surfaces to be devoid of fluid, being either soaked into the articular cartilage or pooled at the lowest point in the joint, by gravity. Larger proteins are, however, adhered to the surface by chemical attraction, providing a constant source of "boundary" lubrication [12]. The extracellular matrix and proteoglycans that form the fibrous basis of the tissue retain synovial fluid near the tissue surface during periods of boundary lubrication such that, on joint loading, it is squeezed into the intra-articular space. This fluid is then drawn in between the two cartilage surfaces, seemingly enabling the femoral condyles to "aqua-plane" over the tibial plateau. This is currently the main theory as to how the synovial joint achieves such little wear and remains healthy for so long; however, this concept cannot be proven due to an inability to visualize synovial fluid via dynamic imaging.

Quite how fluid is drawn between the two opposing surfaces with the intra-articular space remains unknown, defying existing mechanical engineering theories. Mathematical and experimental analyses demonstrate that generation of a "fluid film" is dependent on bearing surface stiffness, smoothness, relative rotational velocity, and contact load. These rules mean that mechanical applications demanding low wear typically use metallic or ceramic bearing materials. When these concepts are replicated in prostheses, longevity rarely extends beyond 20 years and so highlights the natural tissue's remarkable efficacy.

Biomechanical engineers continue to try and understand how the natural joint is so well lubricated. Explants of articular cartilage have been slid against an array of standardized surfaces, enabling cross-comparison of articular cartilage frictional characteristics versus known materials. This has led to the adoption and modification of established mathematical equations to

analytically describe cartilage behavior; however, no models are able to fully explain how a synovial joint can generate highly effective lubrication at very low translation speeds and relatively high compressive loads. Engineers have proposed methods of fluid film generation including the "squeeze film," "boosted," and "weeping" systems, to try and explain how optimal lubrication is achieved in sub-optimal conditions. One of the most recent ideas, of a micro-elastohydrodynamic lubrication system, proposes that the cartilage surface is flattened under loading, making it smoother and so more likely to generate a fluid film. Engineers have now created a mathematical solution that can be considered in the design of replacement joints [13].

23.7 Biomechanical Causes of Knee Degeneration

Osteoarthritis (OA) is the predominant outcome from knee degeneration, with lifestyle and genetics both underlying risk factors. OA is now recognized as a joint disease and is a common cause of disability in people over 65 years. It can lead to the development of bony spurs and changes to the tensile behavior of the ligaments and tendons, changing the articular cartilage's biomechanical environment. An atypical biomechanical environment will cause unequal distribution of loading through the knee, with increased unicompartmental loading thought to be one cause of accelerated joint degeneration [14].

The external knee adduction moment distributes 60–80% of total intrinsic knee compressive load to the medial tibiofemoral compartment, with the lower limb mechanical axis accounting for 50% variation, emphasizing the need for dynamic evaluation. Walking with larger knee adduction moments can increase medial compartment pressure and reduce medial joint space, although this does not appear to cause reduced cartilage thickness, potentially because of deformation in soft structures such as menisci.

While biomechanical factors are also likely to directly contribute to the articular cartilage pathogenesis, the effect of such variables on joint morphology remains unknown. Until recently, biomechanical studies typically considered cross-sectional, rather than longitudinal, knee OA, making differentiation of factors that cause, or result from the disease, difficult. Increased knee loading is known to increase bone mineral density, though little is understood about cartilage response to repetitive, altered load. There is, however, emerging evidence that cartilage volume will become a useful measure in OA pathogenesis, with studies needing to capture the response of human tissues such as hyaline cartilage, to controlled biomechanical variables. The relation between muscle weakness and knee OA is also becoming better understood. Longitudinal studies anecdotally report quadriceps weakness as a feature common to knee OA presentation and degeneration. Baseline knee extensor strength has been shown to be lower in women without radiographic knee OA at the initial examination who later developed OA changes, compared with unaffected women. It may be that weak quadriceps reduce the net extensor moment, which may help to counteract the lateral knee joint opening and medial compression that would occur if the knee adduction moment acted as an unopposed force. Value may be gained by investigating the association between quadriceps strength and the knee adduction moment during gait.

Knee joint laxity, defined as displacement or rotation of the tibia with respect to the femur, is another biomechanical variable argued to contribute to OA pathogenesis. Varus-valgus laxity is reportedly greater in the unaffected knees of patients with unilateral OA than in healthy control subjects, suggesting that knee joint laxity may predispose to disease. It has also been shown that varus and valgus alignment of the lower limb is associated with the progression of medial and lateral compartment knee OA, as determined by joint space narrowing and deterioration of physical function. Moreover, changes resulting from the relation between alignment and disease pro-

gression can be detected after only 18 months of observation. This suggests that over a relatively short intervention time frame, the correction of biomechanical variables in people with established knee OA may delay the progression of disease. Earlier results, however, suggested that women with reduced quadriceps strength have a greater risk of developing knee OA, meaning greater quadriceps baseline strength may be associated with increased OA progression in malaligned and lax knees. Although these results infer that strong quadriceps reduce the risk of developing knee OA, they also suggest that strong quadriceps are a risk factor for the progression of disease in people with malaligned and lax arthritic knees.

Longitudinal studies in normal subjects are still required to determine whether biomechanical variables, such as the knee adduction moment, predate the onset of OA or occur after presentation. Other studies in subjects with OA will be required to clarify the role of biomechanical variables in disease progression, to identify potentially modifiable factors to alter the course of the disease.

23.8 Biomechanics of Implant Design

Designing total joint replacement systems that exhibit both compliance with clinical need and longevity remains a technical challenge. Sir John Charnley was innovating initial joint replacement systems in the 1950s, selecting metallic surfaces to replicate the nuanced attributes of the articular cartilage. Seventy years thence, designers have trialed various systems—including those with stiffnesses more comparable to the soft tissues, though have reverted to adopting traditional engineering materials. Key to the design process is an ability to replicate the generation of the fluid film lubrication layer that separates the two articulating surfaces on translation, minimizing abrasion and so maximizing longevity. While fluid film generation in the healthy joint remains poorly understood, achieving this between two engineering materials requires a high sliding speed, although this typically necessitates an increased sliding distance and so wear. Materials are also required to be hard and smooth, minimizing the layer thickness required to achieve complete separation. Positive conditions are also achieved by having a relatively low contact load, though there is little that can be designed into new components to markedly change these parameters (i.e., this is predominantly a factor of body mass).

The main opportunity for improvement is to focus on achieving high-quality components, as only a small number of materials are available to designers, due to the need for biocompatibility. Efforts to increase the resilience of materials have been successful, with cross-linked ultra-high molecular weight polyethylene producing fewer wear particles than its predecessor. Wear particles are known to accelerate further wear, becoming sandwiched between the two surfaces, like grit. Other improvements have focused on enhancing the implant surface smoothness, to reduce film thickness requirements.

Emerging techniques relate to patient-specific systems, enabling engineers to best match the implant design to an individual's anatomy and so providing opportunity to minimize peak load; however, such systems are more expensive than those 'off the shelf', meaning that they are not yet commonplace. Systems that enable the use of commercially available arthroplasty, though fitted using patient-specific instrumentation, are becoming more popular, potentially offering a more favorable balance between increased longevity and increased cost.

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