

# **PATIENT SPECIFIC KNEE MODEL FOR POST AND PREOPERATIVE SIMULATION AND ANALYSIS**

by

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**PATIENT SPECIFIC KNEE MODEL FOR POST AND  
PREOPERATIVE SIMULATION AND ANALYSIS**

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## ACADEMIC ETHICS AND INTEGRITY STATEMENT

I, Shavkat Nadir Kuchimov, hereby certify that I am aware of the Academic Ethics and Integrity Policy issued by the Council of Higher Education (YÖK) and I fully acknowledge all the consequences due to its violation by plagiarism or any other way.

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## ABSTRACT

### PATIENT SPECIFIC KNEE MODEL FOR POST AND PREOPERATIVE SIMULATION AND ANALYSIS

In this study, we propose a multibody tibiofemoral dynamic model to simulate passive knee motion with the intention of predicting surgical interventions. The knee is a bi-articular joint containing two compartments, tibiofemoral and patellofemoral. Patellofemoral articulation has an important function during active loading of the quadriceps muscle, while tibiofemoral articulation is always involved during both active and passive knee movements. The study focuses on passive knee flexion by proposing a multibody dynamic model of the tibiofemoral joint that incorporates ligaments and contact surfaces. The presence of well-established clinical procedures and publicly available data repositories involving passive knee flexion enables model validation and supports clinical integration. An anatomical knee model is developed, including bones, cartilage, and fifteen ligaments. The ligaments' tensions, the cartilage, and bone contact forces constitute the overall internal force interactions. Passive knee flexion is induced in the form of a joint motion function. It repetitively induced in both the cadaver and the model, resulting in five joint movements: two rotational and three translational. Pearson's correlation resulted in a positive and strong association between the experimental and simulated joint motions ( $r > 0.89$ ). The resulting pairwise linear regression analysis indicates the estimations are statistically significant ( $p < 0.05$ ) for the clinically defined abduction-adduction, internal-external, medial-lateral, anterior-posterior, and superior-inferior motions. The similarity for internal-external rotation has the highest correlation with  $R^2 = 0.994$ . Knee laxity assessment results are consistent with experimental data and literature. The simulated anterior drawer test showed  $> 5\text{mm}$  tibial translation for ACL rupture and  $< 5\text{mm}$  for an intact ACL.

**Keywords:** Tibiofemoral joint model, Knee contact surfaces, Patient-specific ligament forces, Patient-specific ligament attachments, Surgical planning.

## ÖZET

### HASTAYA ÖZGÜ DİZ MODELİ İLE AMELİYAT ÖNCESİ VE SONRASI SİMÜLASYON VE ANALİZİ

Bu çalışmada, cerrahi girişimi öngörmek amacıyla çoklu cisim dinamik tibiofemoral model tasarladık. Diz, tibiofemoral ve patellofemoral olmak üzere iki bölmeli bi-artiküler eklemdir. Patellofemoral eklem kuadriseps kasının aktif kasılması sırasında önemli bir işlev üstlenirken, tibiofemoral eklem hem aktif hem de pasif hareketleri sırasında işlev üstlenmektedir. Bağlar ve temas yüzeylerinden oluşan tibiofemoral eklem modelimizde pasif diz fleksiyon hareketine odaklandık. Ayrıca, iyi yapılandırılmış klinik prosedürlerin varlığı ve pasif diz fleksiyonu içeren verilerin bulunması doğrulama ve klinik entegrasyon aşamasında gerekli olan araçları sağladı. Kemikler, kıkırdaklar ve 15 bağ içeren anatomik diz eklem modeli tasarladık. Bağların gerilimleri, kıkırdak ve kemik temas kuvvetleri gibi iç kuvvet etkileşimleri matematiksel modellerle tasarlandı. Eklem fonksiyonu olarak pasif diz fleksiyonu kullanıldı. Pasif diz fleksiyonu, tanımlı gereği dize sadece fleksiyon-ekstansiyon hareketi uygulanır. Bu hareket hem kadavrada hem de modelde periyodik olarak uygulandı. Hareket neticesinde beklenen beş eklem hareket tepkisi, ikisi açısal ve üçü düzlemsel olmak üzere ortaya çıktı. Pearson korelasyonu, deneysel ve benzetim pasif diz fleksiyonları arasında pozitif ve güçlü bir ilişki ile sonuçlandı ( $r > 0.89$ ). İkili karşılaştırma ile elde edilen lineer regresyon analizi, klinik olarak tanımlanan abduksiyon-adduksiyon, içe-dışa dönme, mediyo-lateral, öne-arkaya ve yukarı-aşağı diz hareketlerin istatistiksel benzerliğinin anlamlı olduğunu ( $p < 0.05$ ) gösterdi. İçe-dışa dönme benzerliği  $R^2 = 0.994$  sayısı ile en yüksek korelasyona ulaştı. Diz laksite değerlendirme sonuçları, deneysel veriler ve literatürle uyumludur. Ön çekme testi benzetimi, ön çapraz bağ (ACL) yırtığında  $> 5$  mm, sağlam ACL durumunda ise  $< 5$  mm tibial öteleme ile sonuçlandı.

**Anahtar Sözcükler:** Tibiofemoral eklem modeli, Diz eklemi temas yüzeyleri, Hastaya özel bağ kuvvetleri, Hastaya özel bağ yapışma konum-ları, Cerrahi planlama.

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## LIST OF SYMBOLS

|                |                                                              |
|----------------|--------------------------------------------------------------|
| $\mathbf{X}_F$ | x-component of the Femoral Coordinate System                 |
| $\mathbf{Y}_F$ | y-component of the Femoral Coordinate System                 |
| $\mathbf{Z}_F$ | z-component of the Femoral Coordinate System                 |
| $\mathbf{X}_T$ | x-component of the Tibial Coordinate System                  |
| $\mathbf{Y}_T$ | y-component of the Tibial Coordinate System                  |
| $\mathbf{Z}_T$ | z-component of the Tibial Coordinate System                  |
| $\alpha$       | Joint rotation about femoral $\mathbf{X}_F$ fixed axis       |
| $\beta$        | Joint rotation about floating axis $\mathbf{F}$              |
| $\alpha$       | Joint rotation about tibial $\mathbf{Z}_T$ fixed axis        |
| $S_1$          | Joint translation along femoral $\mathbf{X}_F$ fixed axis    |
| $S_2$          | Joint translation along floating axis $\mathbf{F}$           |
| $S_3$          | Joint translation along $\mathbf{Z}_T$ fixed axis            |
| $q_1$          | Clinical translation along femoral $\mathbf{X}_F$ fixed axis |
| $q_2$          | Clinical translation along floating axis $\mathbf{F}$        |
| $q_3$          | Clinical translation along $\mathbf{Z}_T$ fixed axis         |
| $a_f$          | represents point a on femur                                  |
| $a_t$          | represents point a on tibia                                  |
| $b_f$          | represents point b on femur                                  |
| $b_t$          | represents point b on tibia                                  |
| $c_f$          | represents point c on femur                                  |
| $c_t$          | represents point c on tibia                                  |
| $d_f$          | represents point d on femur                                  |
| $d_t$          | represents point d on tibia                                  |
| $e_f$          | represents point e on femur                                  |
| $e_t$          | represents point e on tibia                                  |
| $f_f$          | represents point f on femur                                  |
| $f_t$          | represents point f on tibia                                  |
| $g_f$          | represents point g on femur                                  |

|                   |                                                   |
|-------------------|---------------------------------------------------|
| $g_t$             | represents point g on tibia                       |
| $h_f$             | represents point h on femur                       |
| $h_t$             | represents point h on tibia                       |
| $j_f$             | represents point j on femur                       |
| $j_t$             | represents point j on tibia                       |
| $k_f$             | represents point k on femur                       |
| $k_t$             | represents point k on tibia                       |
| $l_f$             | represents point l on femur                       |
| $l_t$             | represents point l on tibia                       |
| $m_f$             | represents point m on femur                       |
| $m_t$             | represents point m on tibia                       |
| $F_i$             | Magnitude of force generated by i'th ligament     |
| $K_{Qi}$          | quadratic stiffness coefficient of i'th ligament  |
| $K_{Li}$          | linear stiffness coefficient of i'th ligament     |
| $\epsilon_i$      | the strain of i'th ligament                       |
| $\epsilon_{1i}$   | i'th ligament's toe to linear transition constant |
| $L_i$             | i'th ligament's instantenous length               |
| $L_{0i}$          | i'th ligament's unstrained length                 |
| $L_{ei}$          | i'th ligament's full-extension length             |
| $e_i$             | the expansion constant for i'th element           |
| $F_c$             | contacting body force                             |
| $\delta$          | the penetration depth                             |
| $k$               | the contact stiffness coefficient                 |
| $p$               | the materials' force deformation properties       |
| $C$               | the damping coefficient                           |
| $h$               | a sigmoid-like step function with a unity gain    |
| $\delta_c$        | a step function transition interval               |
| $F_{ALL}$         | Magnitude of force generated ALL                  |
| $K_{QALL}$        | quadratic stiffness coefficient of ALL            |
| $K_{Li}$          | linear stiffness coefficient of ALL               |
| $\epsilon_{1ALL}$ | ALL's toe to linear transition constant           |

|            |                                             |
|------------|---------------------------------------------|
| $L_{0ALL}$ | ALL's unstrained length                     |
| $y$        | control data for linear regression model    |
| $x$        | comparison data for linear regression model |
| $b_0$      | intercept or comparison data offset         |
| $b_1$      | slope of the control vs comparison data     |



## LIST OF ABBREVIATIONS

|        |                                                      |
|--------|------------------------------------------------------|
| ACL    | Anterior Cruciate Ligament                           |
| MCL    | Medial Collateral Ligament                           |
| CF     | Contact Forces                                       |
| LF     | Ligament Forces                                      |
| JCS    | Joint Coordinate System                              |
| ISB    | International Society of Biomechanics                |
| CT     | Computed Tomography                                  |
| MR     | Magnetic Resonance                                   |
| Lepi   | Lateral epicondyle                                   |
| Mepi   | Medial epicondyle                                    |
| Mppc   | the most posterior point on Medial condyle           |
| Lppc   | the most posterior point on Lateral condyle          |
| Chip   | Center of hip joint                                  |
| Ofem   | Origin of femur                                      |
| Mctc   | Medial tibial condyle                                |
| Lctc   | Lateral tibial condyle                               |
| Otib   | Origin of tibia                                      |
| Cank   | Center of ankle joint                                |
| NK     | Natural Knee                                         |
| CM     | Computational Model                                  |
| MRI    | Magnetic Resonance Image                             |
| MIMICS | Materialise Interactive Medical Image Control System |
| ADAMS  | Automatic Dynamic Analysis of Mechanical Systems     |
| MATLAB | MATrix LABoratory                                    |
| 3D     | three-dimesion                                       |
| T2     | Transverse relaxation time                           |
| DICOM  | Digital Imaging and Communications in Medicine       |
| USG    | Ultrasound Sonography                                |

|      |                                                      |
|------|------------------------------------------------------|
| STL  | Stereo Lithography                                   |
| PCL  | Posterior Cruciate Ligament                          |
| LCL  | Lateral Collateral Ligament                          |
| AM   | Anterior-Medial                                      |
| PL   | Posterior-Lateral                                    |
| aACL | anterior bundle of Anterior Cruciate Ligament        |
| pACL | posterior fibers of Anterior Cruciate Ligament       |
| ALB  | Antero Lateral Bundle                                |
| aPCL | Anterior bundle of Posterior Cruciate Ligament       |
| PMB  | Postero Medial Bundle                                |
| pPCL | Posterior fibers of Posterior Cruciate Ligament      |
| sMCL | supercial Medial Collateral Ligament                 |
| dMCL | deep Medial Collateral Ligament                      |
| MFL  | Menisco Femoral Ligament                             |
| MTL  | Menisco Tibial Ligament                              |
| aMCL | anterior bundle of Medial Collateral Ligament        |
| pMCL | posterior bundle of the Medial Collateral Ligament   |
| pMCL | a posterior bundle of the Medial Collateral Ligament |
| oMCL | an oblique bundle of the Medial Collateral Ligament  |
| dMCL | a deep bundle of the Medial Collateral Ligament      |
| iMCL | inferior fiber of Medial Collateral Ligament         |
| POL  | Posterior Oblique Ligament                           |
| TCL  | Tibial Collateral Ligament                           |
| MFE  | Medial Femoral Epicondyle                            |
| AT   | Adductor Tubercle                                    |
| MGT  | Medial Gastrocnemius Tubercle                        |
| GT   | Gastrocnemius Tubercle                               |
| LGT  | Lateral Gastrocnemius Tendon                         |
| PLT  | Popliteus Tendon                                     |
| FCL  | Fibular Collateral Ligament                          |
| OPL  | Oblique Popliteal Ligament                           |

|        |                                           |
|--------|-------------------------------------------|
| APL    | Arcuate Popliteal Ligament                |
| FBL    | Fabella Fibular Ligament                  |
| lCAP   | Posterior lateral CAPsule                 |
| mCAP   | Posterior medial CAPsule                  |
| ALL    | Antero Lateral Ligament                   |
| ALS    | Antero Lateral Structure                  |
| DoF    | Degrees of Freedom                        |
| CRT    | Clinical Rotations and Translations       |
| FE     | Flexion-Extension                         |
| FE     | Abduction-Adduction                       |
| IE     | Internal-External                         |
| ML     | Medial-Lateral                            |
| AP     | Anterior-Posterior                        |
| SI     | Superior-Inferior                         |
| FM-MT  | Fixed Femur - Moving Tibia                |
| HF-PMT | Hinged Femur - Perpendicular Moving Tibia |
| GCS    | Global Coordinate System                  |
| LCS    | Local Coordinate System                   |
| RMSE   | Root Mean Squared Error                   |
| PKF    | Passive Knee Flexion                      |
| 2D     | Two-dimensional                           |
| F      | Femur                                     |
| T      | Tibia                                     |
| P      | Patella                                   |
| TF     | Tibio - Femoral                           |
| PF     | Patello - Femoral                         |
| FEI    | Finite Element                            |
| Def    | Deformable                                |
| X      | property is absent                        |
| RoM    | Range of Motion                           |
| PLS    | Postero-Lateral Structure                 |

|     |                           |
|-----|---------------------------|
| AL  | Arcuate Ligament          |
| PFL | Popliteo-Fibular Ligament |
| PT  | Popliteus Tendon          |



# 1. INTRODUCTION

Human physical activity is highly correlated with the health. Walking is the basic and necessary activity for human. There are recent studies on walking steps and corresponding daily frequency. Children and adolescents take 11,000 to 15,000 and 10,000 to 11,700 steps per day, respectively, corresponding to 60 minutes of activity for both groups [21]. In contrast, the walking profile for healthy older adults ranges from 2000 to 9000 steps per day, which is equivalent to approximately 150 minutes of daily activity [22]. However, walking activities are sustained only if the individual is healthy and free of lower joint pain. Among all lower extremity joints—hip, knee, and ankle—the knee plays a unique role. Due to its functions in weight-bearing, movement, and load transmission, the prevalence of knee disorders is significant. Knee disorders are the most common lower extremity issues, accounting for 30.3% of cases [23, 24, 25, 26], particularly in sports-related injuries and older adults with arthritis. Studies indicate that knee pain affects 25% of adults, with its prevalence increasing by nearly 65% over the past two decades [27, 28, 29].

The knee joint plays a critical role by absorbing the initial impact during walking. This periodic impact can lead to eventual overuse injuries. Moreover, the knee undergoes rapid load alterations and forces during each step, making it a high-risk area for damage. The knee cartilage is susceptible to wear and degeneration over time due to repetitive stress and high loads, resulting cartilage degeneration. Osteoarthritis commonly affects the knee because of the joints load-bearing function and repetitive use, especially in older individuals [30, 31, 32]. The knee is positioned between the long levers of the femur and tibia, which amplifies mechanical forces across the joint. Misalignment of the leg bones or abnormalities in gait can significantly increase stress on the knee. Consequently, the knee is particularly susceptible to injuries such as ligament tears—including the Anterior Cruciate Ligament (ACL) and Medial Collateral Ligament (MCL)—meniscal tears, and patellofemoral pain syndrome [29, 33, 34, 35, 36]. These injuries often occur when excessive forces or abnormal motions place stress on

the joint tissues beyond their physiological limits [37, 38].

The knee is the primary weight-bearing joint during standing and dynamic activities. It directly supports a significant portion of the body's weight, especially during high-impact activities like running or stair climbing. Unlike the hip, which is cushioned by large muscle groups, the knee relies more heavily on cartilage, menisci and ligaments for shock absorption and stability, which are more prone to injury and wear. The knee has a role of hinge like joint but also has a degree of rotational and translational motions, making it more vulnerable to mechanical stresses from different directions. It has to balance flexion and extension and withstand not only frontal and transverse rotational but also all-direction translational forces, especially during activities that involve sudden directional changes (e.g., running) or during uneven loading (e.g., stair ascending) [39]. The ligaments, tendons, and cartilage in the knee are often under strain to stabilize and align the joint under complex loading conditions, leading to higher susceptibility to wear and tear.

The human knee joint is a complex structure bearing the whole body during various activities resulting in huge local force and torque generations. Hence, it is one of the most frequently injured and degenerating joints. Conventional intervention has numerous surgical procedures on treating injuries, diseases and anomalies of the joint [40]. The main goal of these surgical interventions is reestablishment of physiologically typical loading environment for the knee joint. Previous studies on post surgical knee anomalies reported that joint motions were not consistent. They found altered joint kinematics caused by abnormal ligament loadings may change the contact path and stress within the joint, hence predispose the knee to the degeneration and pain, even for the long-term development of osteoarthritis cite [41].

An in-depth understanding of joint dynamics helps to comprehend injury mechanics, assessment of tissue loading, and consequences of surgical intervention. The resisting force components driving passive knee motion are due to rigid body contact interactions and ligament forces. The sound knee joint has a non-homogeneous oval shape and smooth surface at the distal end of the femur consisting of medial and

lateral condyles. It has a varying curvature radius of instantaneous contact points at the sagittal plane and posterior cross sections as illustrated in Figure 1.1a and b, first introduced by Kapandji [1]. In the sagittal plane, hypothetical contact circles have a lower radius at higher flexion angles than near full extension angles [1, 2]. In the frontal plane, the contact surface of the medial condyle is a circular arc and the lateral condyle surface is flattened at full extension (Figure 1.1c) [3]. Posterior cross sections of condyles are circular and their radii are not uniform for the range of the knee flexion motion as shown in Figure 1.1d. Femoral condyles are mechanically coupled with the proximal head of the tibia containing plateaus which have non-homogeneous medial concave and lateral saddle shapes [42]. Contact of these surfaces, made of bones and cartilage, produces Contact Forces (CF).

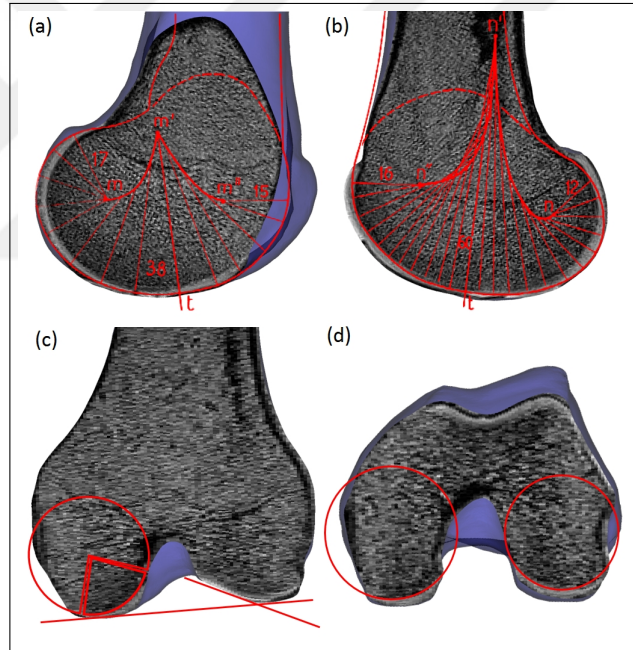


**Figure 1.1** Contact surfaces of femoral condyles are depicted through cross sections of the distal femur. Diagrams of varying radii of curvatures are superimposed on sagittal sections of the medial (a) and lateral (b) condyles segmented from MRI [1, 2]. The frontal section (c) of the medial condyle(left) has a circular arc whereas the lateral one(right) has a flat surface at full extension. The transverse section (d) of both medial (left) and lateral (right) condyles posteriorly have a circular shape along higher knee flexion [3].

Anatomically, ligaments and capsule have attachment sites on the femur, tibia and fibula. Under tension, these ligamentous structures become a source of passive Ligament Force (LF). The interaction of CF and LF together defines the main force structure. Therefore, a generic comprehensive knee model must possess all these force

components customized to a subject.

The uniaxial hinge-like behavior of the knee joint in quadruped mammals is constrained to one degree of freedom, allowing only flexion-extension motion [43]. Unlike in mammals, the human knee is one of the most complex joints and exhibits multiple degrees of freedom. In addition to the obvious and well-known bending, or flexion-extension movement, the knee allows angular motions such as internal-external rotation about the long axis of the bones and abduction-adduction of the femur or tibia relative to the body's centerline. Furthermore, the knee exhibits translational motions, including medial-lateral, anterior-posterior, and superior-inferior displacements. These translational motions are named based on the displacement of one segment relative to the other.

Although not a formal rule, physicians conventionally describe joint movements with the proximal segment (closer to the hypothetical body center) as fixed and the distal segment (farther from the center) as moving. For the knee joint, the proximal segment is the femur, while the distal segment is the tibia. Thus, when a physician describes knee joint movement, they typically consider the tibia moving relative to the femur. For instance, knee flexion, abduction, and external rotation refer to the tibia bending backward in the sagittal plane, angling away from the body's centerline in the frontal plane, and rotating clockwise for the left knee and counterclockwise for the right knee in the transverse plane. Similarly, medial, anterior, and superior translations of the knee indicate the tibia moving toward the body's centerline along the sagittal axis, forward along the frontal axis, and upward (toward the head) along the transverse axis.

While there are no universally strict rules for defining joint motions, the International Society of Biomechanics (ISB) has recommended standards for the Joint Coordinate System (JCS) to enhance communication among researchers and clinicians. These standards provide a framework for consistent and precise descriptions of joint movements, facilitating interdisciplinary collaboration and improving the reproducibility of studies. However, the most standardized and widely recognized definition of knee joint movement was established by Grood and Suntay in their seminal 1983 study [44]. Their work

introduced the concept of the Joint Coordinate System for the human knee, which has since become a cornerstone in biomechanical research. Grood and Suntay's method defines joint movement based on three orthogonal axes: the flexion-extension axis, the longitudinal axis of the tibia, and a floating axis that varies depending on the movement. This coordinate system effectively separates rotational and translational motions, allowing for a clearer representation of the knee's complex mechanics. By anchoring the coordinate system to anatomical landmarks, their approach provides a standardized way to measure and describe motions such as flexion-extension, internal-external rotation, and abduction-adduction, as well as translational movements along the medial-lateral, anterior-posterior, and superior-inferior axes. This framework is particularly beneficial in both *in vivo* and *in vitro* studies of knee joint motion, where testbeds are designed to assess the movement of the femur, tibia, or both segments. Regardless of the specific experimental setup, Grood and Suntay's definition ensures that the motion of each segment is described relative to a consistent reference entity, minimizing ambiguity and enhancing comparability between studies. The adoption of their standardized system, alongside ISB recommendations, has significantly advanced our understanding of knee joint biomechanics and continues to serve as a foundational reference in the field.

Clinically, joint axes are often used as reference entities, with conventions and nomenclature evolving through extensive experience. In this study, the reference coordinate system is fixed to one of the rigid components, which is also aligned with the global coordinate system, as determined by the study's objectives. The terminology used to describe joint motions can be joint-specific or part-specific and is often used interchangeably by physicians. For instance, "genu varum" refers to tibial adduction, while "genu valgum" describes tibial abduction. Here, "varum" and "valgum" are joint-specific terms, whereas "adduction" and "abduction" are part-specific. Adduction refers to the movement of a body part toward the midline, while abduction describes movement away from it. Anatomically, this means the distal part (e.g., the tibia) moves toward the proximal part's midline (e.g., the femoral axis) during adduction and away from it during abduction. Additionally, internal movement of the tibia at the knee joint corresponds to external movement of the femur when the tibia is fixed, and vice

versa.

Proper expression of translational motions along the axes is crucial. For example, during translation along the anterior-posterior axis, the tibia moves forward during an anterior drawer test, whereas the femur moves backward relative to the tibia. The same principle applies to translational movements along the medial-lateral and superior-inferior axes. Therefore, knee joint motion descriptions must clearly specify both the moving segment and the referenced segment.

Several studies have investigated knee passive motion, with the primary goal of creating experimental databases for the passive motion of human knee joints. These databases capture the knee joint's freedom-of-motion characteristics and are also used to enhance and validate computational knee models. Passive knee motion studies mainly focus on the limits or envelopes of motion. During incremental flexion and extension, these studies examine the ranges of freedom-of-motion, such as internal-external rotation, valgus-varus rotation, and anterior-posterior or compressive-distractive joint translations. One of the earlier investigations into the envelope of knee joint passive motion was conducted by Blankevoort et al [4]. A cadaver knee was mounted on a six-degree-of-freedom apparatus (Figure 1.2), where it was exposed to flexion-extension movement, resulting in induced axial forces, anterior-posterior forces, medial-lateral forces, varus-valgus torques, and internal-external torques.

Previous studies generally focus on the contact forces related to the geometry of anatomical surfaces, and the constraint forces associated with connective tissue. The anatomical representation varies depending on the level of detail considered for the knee joint. Several computational models are designed as two-dimensional models since the main motion occurs in the sagittal plane, which also facilitates the use of solvable mathematical expressions [45, 46, 47]. Progression in knee model design leads to employ anthropometric parameters. For instance, Abdel-Rahman and Wilson et al. approximate the femoral condyles with a pair of spherical shells [48, 49]. They fitted spheres to the medial and lateral condyles and used the radii as an anatomical feature. They and Shelburne et al. treated medial and lateral tibial plateau surfaces as

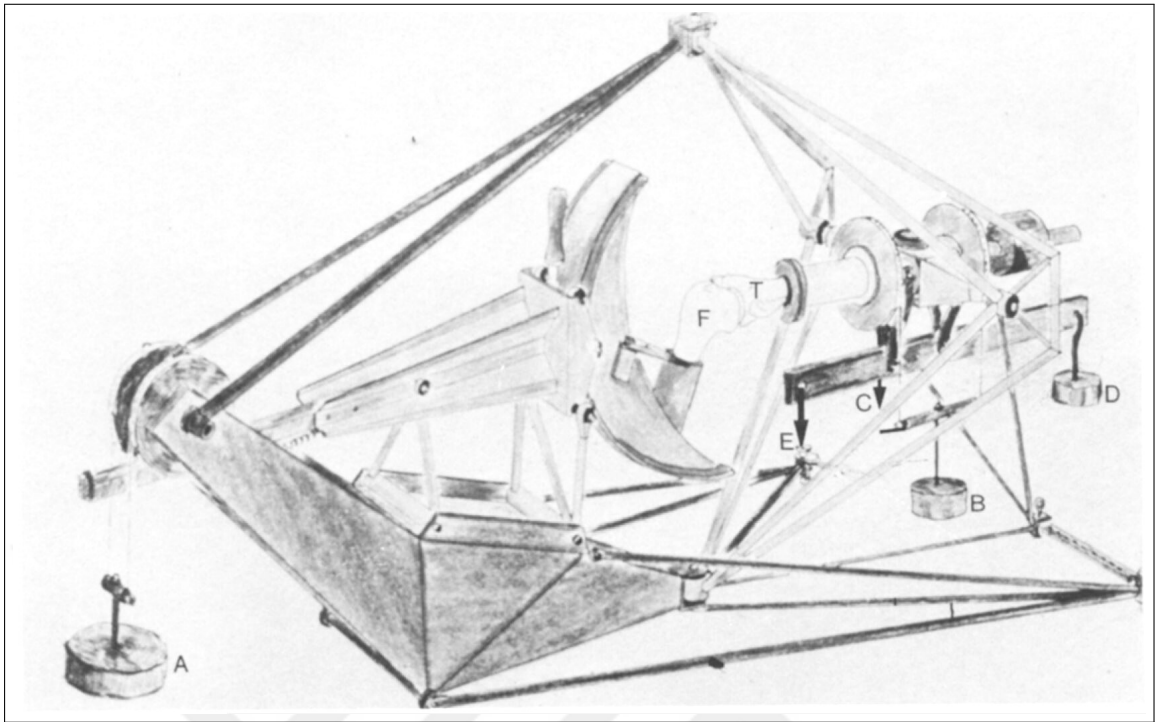


**Figure 1.2** Knee joint apparatus reproduced from [4]. F - femur, T - tibia. Rotations: flexion-extension, varus-valgus, internal-external, and Translations: medial-lateral, anterior-posterior, proximal-distal. A cadaver knee was exposed to flexion-extension movement induced axial forces, anterior-posterior forces, medial-lateral forces, varus-valgus torques, and internal-external torques.

inclined planes, with slopes measured from specimens [48, 49, 47]. Although femoral condyle diameters and tibial plateau slopes contribute to anatomical accuracy, articular surfaces do not have uniform condylar radii nor plateau slopes completely define contact surfaces. Later knee morphology studies showed that femoral condyles have varying radii and tibial plateaus are not necessarily planar [2, 3].

In a study by Akalan et al. the tibia with flat surfaces constrained by isometric ligaments was moved on the fixed femur, resulting in contact traces resembling anatomical condyle contours [50]. However, these curves do not represent the entire surface morphology of the condyles, thus limiting the model's ability to calculate internal-external, abduction-adduction rotations, and lateral-medial translation motions that are out of the contact pattern.

Some studies employ scanned or probed surface geometries of specimen knee articulation, Contour or surface functions are derived in a way to simplify the calculations

of geometric and dynamic equations. In Wismans et al.’s model, generalized surfaces of femoral and tibial condyles are approximated by higher-order mathematical polynomials. However abduction-adduction knee rotation cannot be studied due to assumptions of persisting permanent contacts on medial and lateral condyles [51].

Gerus et al. showed that subject-specific tibiofemoral joint geometry improves contact force prediction on the medial side compared to generic models [52]. Recent studies utilize Compute Tomography (CT) and Magnetic Resonance (MR) imaging modalities to extract subject-specific contact surfaces at the knee joint [53, 5, 6, 54, 52]. This approach benefits both cadaver and in-vivo studies since dissection and surface probing are no longer needed for knee articulation modeling.

Multibody computational models presented by Guess et al., based on cadaver and subject studies, consider the main interacting force components including menisci [54]. However, knee capsules, unlike the anterolateral ligament, which can vary among human races, are anatomically present in every knee but are often excluded from models. A summary of the knee models is provided in a table in Appendix A.1, which includes details of the models in the literature.

In this study, we propose a multi-body anatomic tibiofemoral model. Unlike existing approaches, we introduce a more generalized model that encompasses a broader range of knee motions. With its comprehensive representation of connective tissues and three-dimensional contact surfaces, the proposed tibiofemoral model allows for patient-specific simulations, capturing the six degrees of freedom joint motion during passive knee flexion.

In Chapter 2, the materials and key mechanical components of the proposed model are detailed in five sections. The first section outlines the rigid body coordinate system and joint axes of the knee. The second section presents the data and tools utilized during the data preparation and analysis phases. The segmentation and landmark identification section explains how bones and cartilage tissue are extracted as shells, and how anatomical points are located to define the rigid body coordinate system

using radiological images. Additionally, the fourth section describes how ligament attachment points and their bundles are registered using the same images. Finally, the construction of the computational model is presented in Section 5.

In Chapter 3, simulated passive flexion data from the computational knee model are compared with benchmark experimental data. The findings of the study are discussed in Chapter 4. Lastly, Chapter 5 presents the discussion and conclusion, where the results of the intact computational model and the ligament-teared computational model are analyzed.

## 1.1 List of publications produced from the thesis

1. Impact of subject's physical properties on joint biomechanics: Hypermobility alters lower extremity biomechanics during knee-bearing activity, S. Kuchimov, M. Özkan, A. Apti, N.E. Akalan, B. S. Akel, K. Hollander, " *The European Society for Movement Analysis for Adults and Children, ESMAC, 2023 Annual Meeting of the European Society, Gait and Posture*, Vol. 3, Supp. 1, Pages S107-S108, Sep. 18–Sept. 23, Athenes, Greece, 2023.
2. A validation of tibio femoral joint model: Quantitative comparison of experimental and model response kinematics, S. Kuchimov, M. Özkan, N. E. Akalan, C. A. Yucesoy, Y. Temelli, *The European Society for Movement Analysis for Adults and Children, ESMAC, 2022 Annual Meeting of the European Society, Gait and Posture*, Vol. 97, Supp. 1, Pages S40-S41, Sep. 19–Sept. 25, Dublin, Ireland, 2022.
3. Multi-body dynamic model of subject specific knee: Utilization of mri images for reconstruction of contact bodies and insertion sites, S. Kuchimov, F. Bilgili, N. E. Akalan, Y. Temelli, E. Abdel-Rahman, M. Özkan, *The European Society for Movement Analysis for Adults and Children, ESMAC, 2021 Annual Meeting of the European Society, Gait and Posture*, Vol. 90, Supp. 1, Pages 131-132, Oct. 7–Oct. 15, Odense, Virtual, 2021.

## 2. MATERIALS AND METHODS

### 2.1 Rigid body coordinate systems and joint axes

A knee joint involves four main bones as rigid body structures: the femur, tibia, patella, and fibula. It is common to treat the tibia and fibula as one rigid body (hereinafter referred as the tibia). We exclude the patella from consideration due to its insignificant effect on passive knee flexion. Therefore, the rigid bodies of the proposed model are referred to as the femur and tibia. The rigid body coordinate frames and joint axes are constructed as described by Grood and Suntay [44], which are relevant to anatomical landmarks and axes. The femoral and tibial cartilages are attached to the distal head of the femur and the proximal head of the tibia.

#### 2.1.1 Cartesian coordinate system of rigid bodies

Rigid body coordinate frames are defined at the distal femoral and proximal tibial heads using anatomical landmarks illustrated as red dots in Figure 2.1. Femoral landmarks include the **L**ateral (*Lepi*) and **M**edial (*Mepi*) **epi**condyles; the most **p**osterior points on the **M**edial (*Mppc*) and **L**ateral (*Lppc*) condyles; the **C**enter of the **hip** ball (*Chip*) at the proximal femoral head; and **f**emoral **O**origin (*Ofem*) the most distal point on the posterior surface of the femur equidistant to *Lepi* and *Mepi*. The vector pointing from the origin *Ofem* to *Chip* is the longitudinal axis of the femur,  $\mathbf{Z}_F$ .  $\mathbf{Y}_F$  is the cross-product of  $\mathbf{Z}_F$  with the unit vector parallel to the vector pointing from *Mppc* to *Lppc*. Finally,  $\mathbf{X}_F$  is the cross-product of  $\mathbf{Z}_F$  and  $\mathbf{Y}_F$  base vectors.

The tibia has four anatomical landmarks as seen in Figure 2.1. Two of them are the center points of the **M**edial (*Mctc*) and **L**ateral tibial condyles (*Lctc*). The **a**nkle **C**enter (*Cank*) is the midpoint between the medial and lateral malleolus at the ankle joint. The **O**origin of the **t**ibial coordinate system (*Otib*) is the midpoint between the

tibial eminences. The tibial mechanical axis is a vector pointing from *Cank* to *Otib*, referred to as the  $\mathbf{Z}_T$ -axis.  $\mathbf{Y}_T$  is obtained by the cross-product of  $\mathbf{Z}_T$  and the unit vector directed from the point *Mctc* to *Lctc*. The third axis,  $\mathbf{X}_T$ , is orthogonal to  $\mathbf{Z}_T$  and  $\mathbf{Y}_T$ .

### 2.1.2 Joint axes

Three base axes of the joint coordinate system are depicted as black arrows in Figure 2.1. The two joint axes are embedded in the femur and tibia, coinciding with the femoral  $\mathbf{X}_F$  and tibial  $\mathbf{Z}_T$  axes, respectively. The third joint axis,  $\mathbf{F}$ , is floating and always orthogonal to both  $\mathbf{X}_F$  and  $\mathbf{Z}_T$ . Rotational motions - flexion-extension ( $\alpha$ ), external-internal ( $\gamma$ ) and abduction-adduction ( $\pi/2-\beta$ ) - occur about the  $\mathbf{X}_F$ ,  $\mathbf{Z}_T$ , and  $\mathbf{F}$  axes, respectively.

The vector pointing from femoral origin (*Ofem*) to tibial origin (*Otib*) is translation vector  $\mathbf{H}$  represented by black thin arrow in Figure 2.1. The components of  $\mathbf{H}$  along joint coordinate axes are joint translations (indicated by dashed black lines with arrowheads), while projections of  $\mathbf{H}$  onto joint axes are clinical translations (represented by dashed colored lines with arrowheads). Although joint translation ( $S_2$ ) always equals clinical drawer ( $q_2$ ) along the floating axis ( $\mathbf{F}$ ), this is not the case for translations along the fixed axis. Lateral-medial ( $S_1$ ) and superior-inferior ( $S_3$ ) joint translations are not equal to clinical lateral-medial trust ( $q_1$ ) and clinical distraction-compression ( $q_3$ ).

In Figure 2.1, the tibia is posteriorly moved to eliminate anterior-posterior translation for illustration purpose of joint and clinical translations. For both rigid bodies, the positive directions are proximally for the  $Z$ -axis, anteriorly for the  $Y$ -axis, laterally for the right limb's  $X$ -axis, and medially for the left limb's  $X$ -axis.

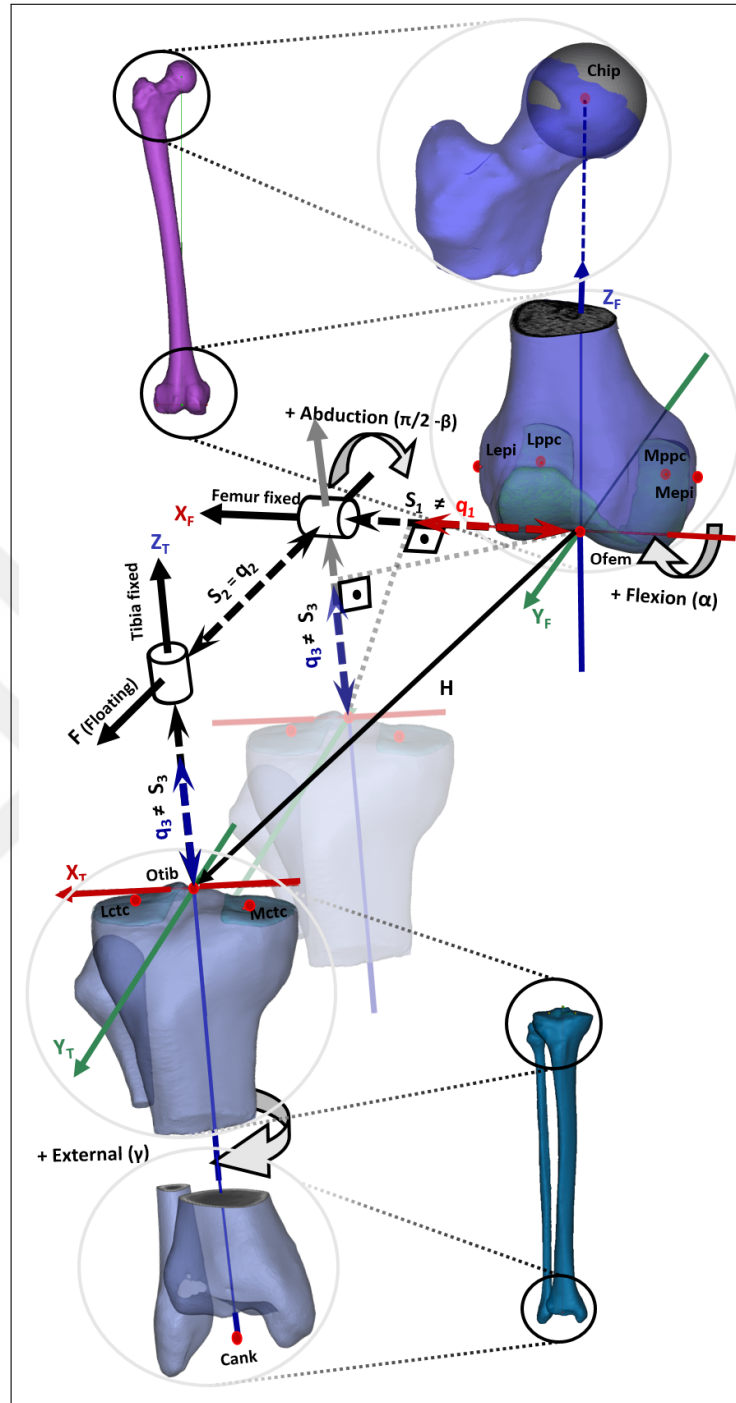


**Figure 2.1** Cartesian coordinate systems are constructed using red dots. Axes are colored red ( $\mathbf{X}$ ), green ( $\mathbf{Y}$ ), and blue ( $\mathbf{Z}$ ) with subscripts F and T denoting femur and tibia, respectively. Black bold arrows represent the three joint coordinate axes used for describing joint rotations and translations. The knee flexion-extension ( $\alpha$ ) occurs about the femur fixed axis ( $\mathbf{X}_F$ ), the tibial internal-external rotation ( $\gamma$ ) occurs about the tibia fixed axis sharing ( $\mathbf{Z}_T$ ), and the abduction-adduction ( $\pi/2 - \beta$ ) occurs about a floating axis ( $\mathbf{F}$ ), which is the common perpendicular to both. The translation vector ( $\mathbf{H}$ ) has lateral-medial ( $S_1$ ), anterior-posterior ( $S_2$ ), and superior-inferior ( $S_3$ ) components, which are referred as joint translations and are shown as dashed black lines with arrowheads.  $S_2$  is always equal to the corresponding clinical translation, the anterior-posterior drawer ( $q_2$ ). If the joint is abducted or adducted,  $S_1$  and  $S_3$  will not be equal to the clinical thrust ( $q_1$ ) and distraction-compression ( $q_3$ ) which are the projections of  $\mathbf{H}$  on the fixed axes. These projections are illustrated on a transparent tibia moved along  $\mathbf{F}$ .

## 2.2 Data and tools

### 2.2.1 Specimen repository

To validate the proposed knee model, we utilized a repository of Natural Knee (NK) data from the Digital Commons at DU, which corresponds to a 70 kg, 183 cm, 44-year-old male cadaver [5, 6]. The experimental dataset includes both CT and MR images. The CT scan data has the following specifications: 356 slices, slice thickness of 0.6 mm, in-plane pixel spacing of 0.4x0.4 mm, in-plane resolution of 512x512 pixels, tube current of 130 mA, and a peak potential of 120 kV. The MR data were acquired using the sequence T2-weighted, True Fast Imaging, 3D, Water Excitation, Sagittal Plane T2trufi3d\_we\_SAG. MR data include 192 slices with a thickness of 0.55 mm, in-plane pixel spacing of 0.53125x0.53125 mm, and an in-plane resolution of 320x320 pixels. A tracking device was employed to collect a set of landmark points, and a fiducial markers, including ligament attachment sites. Marker trajectories were compiled by tracking segment motion resulting of manually applied torques and displacements as depicted in Figure 2.2 [5, 6]. Next, joint kinematics and kinetics were computed for both passive knee flexion and knee laxity tests using the marker trajectories. A common frame of reference was established as illustrated in Figure 2.1. We transformed these computed trajectories into the Computational Model's (CM) frame of reference for comparison purposes. Therefore the repository data serves as the gold standart to verify the results of the proposed model.

The proposed method achieves patient specificity by utilizing the MR images of the subject. Patient-dependent bone and cartilage geometries are segmented from MR images, along with ligament and capsule attachment sites, which are also patient specific. These segmented structures define contact geometries, anatomical landmarks, and rigid bodies in Cartesian space. Ligament and capsule attachment sites are located on the MRI based on previous anatomical studies [8, 10, 11, 12, 13, 14, 15, 16, 17, 18, 55, 56, 57, 58, 59, 60, 61]. By registering landmark sites on both MR and CT images, MRI identified ligament sites can be validated with the actual sites measured on the cadaver.

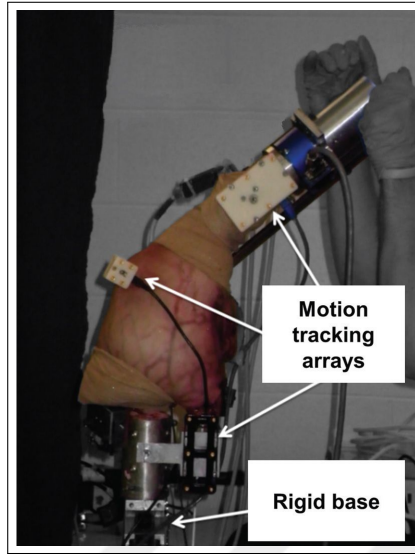


**Figure 2.2** Passive flexion movements applied to the cadaver leg. Experimental data were gathered by applying external force manually (Experimental installation picture is used with the permission of the authors [5, 6]).

The identified landmarks and segmented structures provide the basis for creating a twin model of the cadaver in the simulation environment. The passive knee flexion motion data provided with the repository is used to benchmark the kinematics of our model.

### 2.2.2 Mimics

Materialise Interactive Medical Image Control System (MIMICS) is an advanced medical imaging and engineering software developed by Materialise NV [7]. It is widely used in biomedical field for process and analysis of radiological images obtained by several modalities i.e; X-ray, CT, MR and Ultrasound Sonography (USG). Using those images, it assists to creates 3D models for various applications, including research, clinical, and industrial purposes.

Bone and cartilage segments, anatomical landmarks, and ligament attachment sites are obtained from repository MR images using MIMICS (Materialise NV, Leuven, Belgium). After segmentation, residual islands are removed, and extracted parts are smoothed and remeshed if necessary. Coordinate points for anatomical landmarks,

ligament, and capsule attachment sites are identified and corroborated with the literature [8, 10, 11, 12, 13, 14, 15, 16, 17, 18, 55, 56, 57, 58, 59, 60, 61].

### 2.2.3 Adams

Automatic Dynamic Analysis of Mechanical Systems (ADAMS) is a multibody dynamics simulation tool for modeling the motion and forces of interconnected bodies [62]. It supports engineers to design and optimize mechanical systems by simulating real-world operating conditions virtually, reducing the need for physical prototypes.

In this study, the proposed model is constructed with ADAMS in which bones, cartilage, and ligaments constitute the interconnected bodies while contact forces and ligament strain define the dynamics of the tibiofemoral model. Rigid body coordinate frames for femur and tibia bones are defined using the coordinate points obtained from Mimics software. Six degrees of freedom and relative motions for the knee joint are defined according to an anatomic coordinate system and clinical joint axes [44]. Contact surfaces segmented from femoral and tibial cartilages are fixed to the corresponding rigid bodies.

### 2.2.4 Matlab

MATLAB stands for MATrix LABoratory. It is high-level programming language and interactive environment developed by MathWorks [63]. It is delicate engineering software that enables engineers to have numerical computation, data analysis, development of algorithm, visualization and simulation.

The experimental and computational model-derived data are filtered, processed, parametrized, and compared using MATLAB software. A regression model is utilized to compare similarity in the simulated and experimental kinematics quantitatively.

## 2.3 Segmentation and landmark identification

In this study, we utilize MIMICS [7] to extract bone and cartilage segments and identify anatomical landmarks in three-dimension (3D). MIMICS is medical imaging software that offers useful measuring, editing and segmentation tools. We use segmented bone and cartilage surfaces for computing contact forces. We define ligament attachment sites using anatomical landmarks on the bone structures. MR images we used are obtained from the NK data repository introduced in Section 2.2.

### 2.3.1 Segmentation of knee joint bones and cartilages

For the contact surface corresponding to the articular joint, CT and MRI data from the repository were considered for use [5, 6]. Since the femoral and tibial contacts involved cartilaginous rather than osseous tissues, Magnetic Resonance Images have to be utilized to reconstruct three-dimensional shell of the cartilages. Bone tissue can be segmented from both MR and CT slice. We extract bone from both modalities for consistency comparison of the joint contact surface.

Initially, MR Digital Imaging and Communications in Medicine (DICOM) images are imported into MIMICS to establish orientation in three orthogonal views: coronal, sagittal, and axial, as shown in Figure 2.3a. Subsequently, bone and cartilage masks are identified interactively using the 3D Live Wire process, as illustrated in Figure 2.3b through g. The process begins with user-defined segmentation points on the frontal plane (Figure 2.3b) and following sagittal (Figure 2.3d) plane. Contours between these points are automatically generated using an energy minimization function. In the axial view, contours constructed in the coronal and sagittal planes appear as horizontal (Figure 2.3c) and vertical red lines (Figure 2.3e), respectively. From these contours, edges for bones and cartilage are automatically generated (Figure 2.3f), followed by manual editing to remove artifacts in the axial images (Figure 2.3g).

In the final stage of the 3D Live Wire process, bone and cartilage tissue are

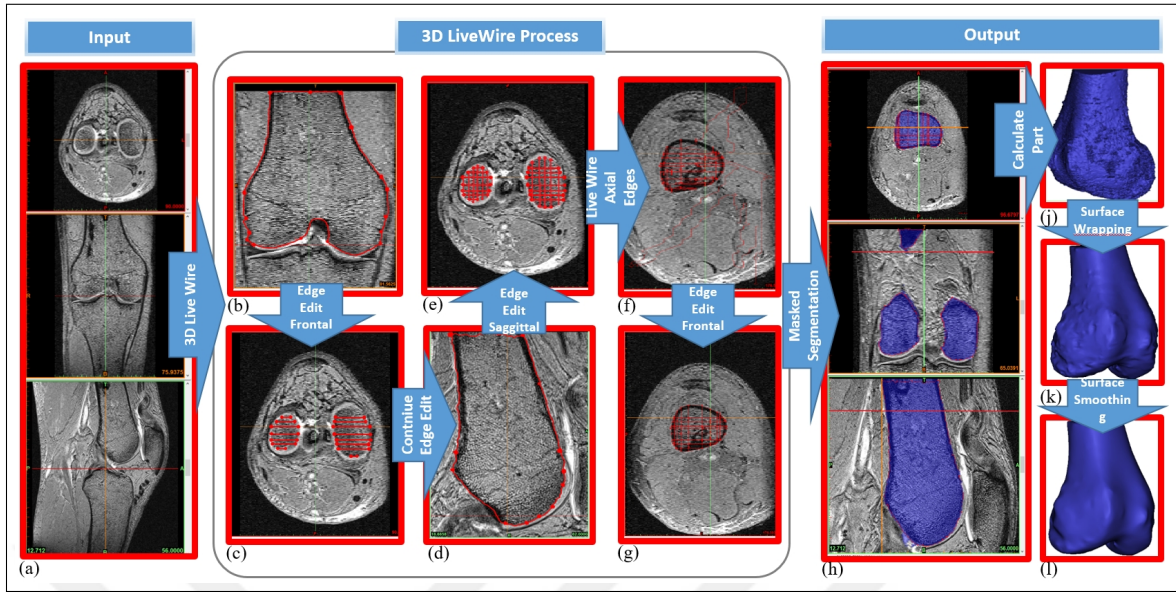


**Figure 2.3** Extraction of femoral bone from MR images. (a) Transverse, frontal and sagittal DICOM images are fed into Mimics[7]. The 3D Live Wire function is utilized to -detect bone surface using sample contours provided on two orthogonal planes, i.e., (b) coronal and (d) sagittal planes. The contours are edited manually to remove of spurious edge points. The contour traces become visible as (c) horizontal and (e) vertical lines on the axial images. (f) Mimics generates bone contours in the third orthogonal plane (axial), (e) which are also edited manually. In Mimics (h) masks are generated from the edges, and (j) the surface object is calculated using masks. (k) Surface wrapping is used to remove spurious holes and bumps. h) A final smoothing yields a more accurate bone surface for computing dynamic contact forces.

segmented in three orthogonal views and for all slices, resulting in blue masks, as illustrated in Figure 2.3h. Using these masks, a 3D object is constructed (Figure 2.3j). The Surface Wrap function in MIMICS is used to remove spurious holes (Figure 2.3k), and the 3D Smooth function is applied to reduce rough or sharp structures, yielding a more realistic bone surface (Figure 2.3l). Redundant meshes are simplified using the Reduce Function included in the software, significantly decreasing the size of the 3D object files. The same procedures are applied to segment bones (patella, fibula, and tibia) as well as cartilages (femoral, patellar, and tibial cartilages). Finally, the segmented shells are converted into 3D objects and exported as Stereo Lithography (STL) files, as shown in Figure 2.4a.

Similarly, knee joint bones are extracted from CT DICOM images using the same procedure, as shown in Figure 2.4b. The results are compared with MRI-segmented bones and found to be consistent. For further illustration, cartilages obtained from

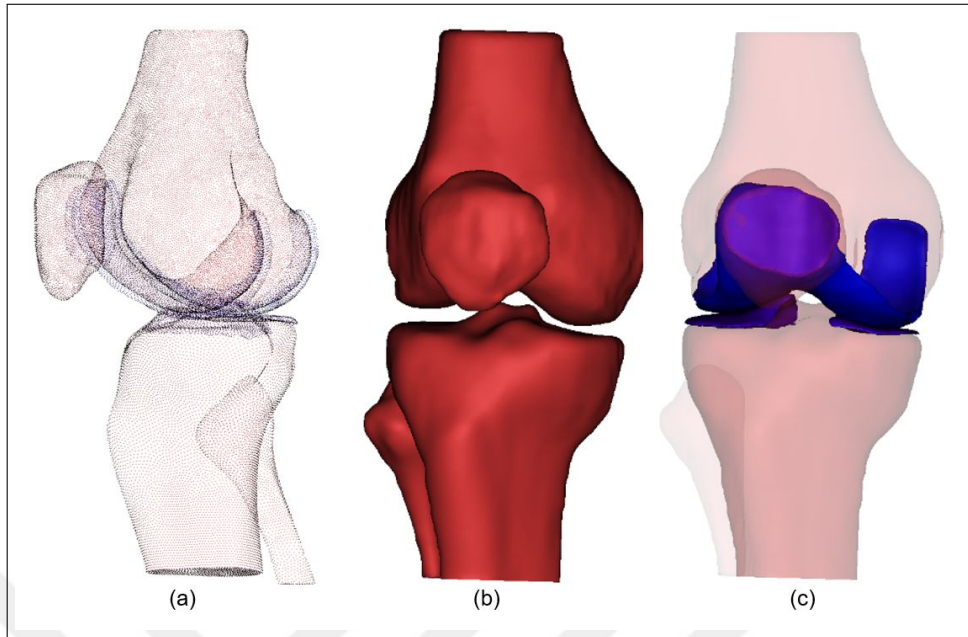


**Figure 2.4** Segmented bones and cartilages of the knee joint. (a) Distal femur, patella, proximal tibia, and fibula segmented using Computed Tomography (CT) images. (b) Femoral, patellar, and tibial cartilages segmented from Magnetic Resonance Imaging (MRI) data and aligned with CT-extracted bones. (c) Bone and cartilage tissues segmented exclusively from MRI slices.

MRI are superimposed onto bones segmented from the CT scan (Figure 2.4c).

## 2.4 Subject specific ligament origins and insertion sites

A better understanding of the functional anatomy and biomechanics of knee structures has been achieved in recent decades. However, these aspects have often been described as qualitative and generic rather than quantitative and specific [12]. To achieve the aim of this study, the attachment sites of ligaments, which determine their lengths, must be tailored to the subject. Nonetheless, advanced knowledge is required to accurately extract these locations.

While a vast number of books and articles on the functional anatomy of the knee offer invaluable insights, there are relatively few quantitative studies focusing on the morphology of bone-to-bone connective tissues associated with the knee. Among these, cadaver studies [8, 12, 13, 16] have provided essential qualitative understandings

of anatomy alongside quantitative measurements of ligament attachment locations, osseous landmarks, and site areas. Despite these contributions, the complex nature of the knee leaves significant areas yet to be explored.

In vitro measurements of specimen parameters provide valuable insights into human anatomy and remain indispensable. However, they are not practical for clinical applications. Theoretically, tools such as CT for identifying bone landmarks and MRI for locating ligament attachment sites can be used to obtain the parameters required for modeling. In practice, it is not convenient to have both scan from a patient. In contrast to specimen studies, research in these fields is often qualitative, with most efforts focused on diagnostic and monitoring purposes.

Few studies have addressed the specific locations of ligament insertions. Victor et al. examined the intra- and inter-observer variability in registering osseous landmarks on CT scans [64]. They determined the anatomic coordinate system, the origin of the coordinate system, and reference points for defining ligament attachment sites based on bone landmarks. Seim et al. attempted to reconstruct ligament and tendon attachment sites using CT images. However, their findings were comparable only for the Posterior Cruciate Ligament (PCL), and the study lacked validation [65].

MRI is the most commonly used modality for imaging soft tissues. Innocenti et al. developed a method to localize ligament insertions on the tibia and femur, specifically for the ACL, PCL, MCL and Lateral Collateral Ligament (LCL). They reported that the procedure was both reproducible and accurate. Furthermore, the results were independent of the operator's proficiency in MRI. Tutorials were noted to improve the operator's understanding of the steps, enhancing accuracy [9]

The research team led by Kevin Shelburne investigated cadaver knees for modeling studies and shared data from their laborious research, referred to as the 'Natural Knee' project [5, 6]. They provided CT images, MRI data, ligament attachment points, and certain landmarks required for calculating the anatomic coordinate system of bones, which were measured using a probe.

We applied Innocenti’s method to one of the samples provided in the ‘Natural Knee’ project. Since the specimen’s point cloud for ligament insertions was not grouped by individual bundles but rather by insertion edges, the comparison between all centroids of the point cloud and the locations obtained using MRI showed deviations for some ligaments. In particular, clustering the bundles of the MCL varied significantly in the literature due to the wide area of its insertion site on the tibia.

The coordinate system defined in Figure 2.1 relies on the number of anatomical landmarks as described in Section 2.1. Coordinate points for the anatomical landmarks, ligament, and capsule attachment sites are manually identified on the MR images within the MIMICS environment with respect to segmented bones. These coordinate points are then cross-referenced with existing literature for validation [8, 10, 11, 12, 13, 14, 15, 16, 55, 56, 57, 17, 58, 59, 60]. For the purposes of this study, the NK data were utilized to develop a subject-specific knee model. Additionally, all discussed methods for locating ligament insertions were applied for consistency and comparison.

### **2.4.1 Anterior Cruciate Ligament**

Generally, an Anterior Cruciate Ligament (ACL) is recognized as not being a simple band of fibers with constant tension. It has been widely accepted as a basis of function separated into Anterior-Medial (AM) or commonly known anterior bundle of Anterior Cruciate Ligament (aACL) and Posterior-Lateral (PL) or posterior fibers of Anterior Cruciate Ligament (pACL) [8]. Investigation performed on 7 cadaveric knees were projected onto the surface of the lateral femoral condyle’s tibial head and medial wall.

The central fibers of the aACL or the AM bundle of the ACL are marked with white dots on the tibia and red dots on the femur as shown in Figure 2.5a, b and Figure 2.6a, b, respectively. Similarly, the pACL or th PL bundle of the ACL are marked with a black dot on the tibia (Figure 2.5a and b) and a green dot on the femur (Figure 2.6a and b). These attachment points were projected onto the dissected femur

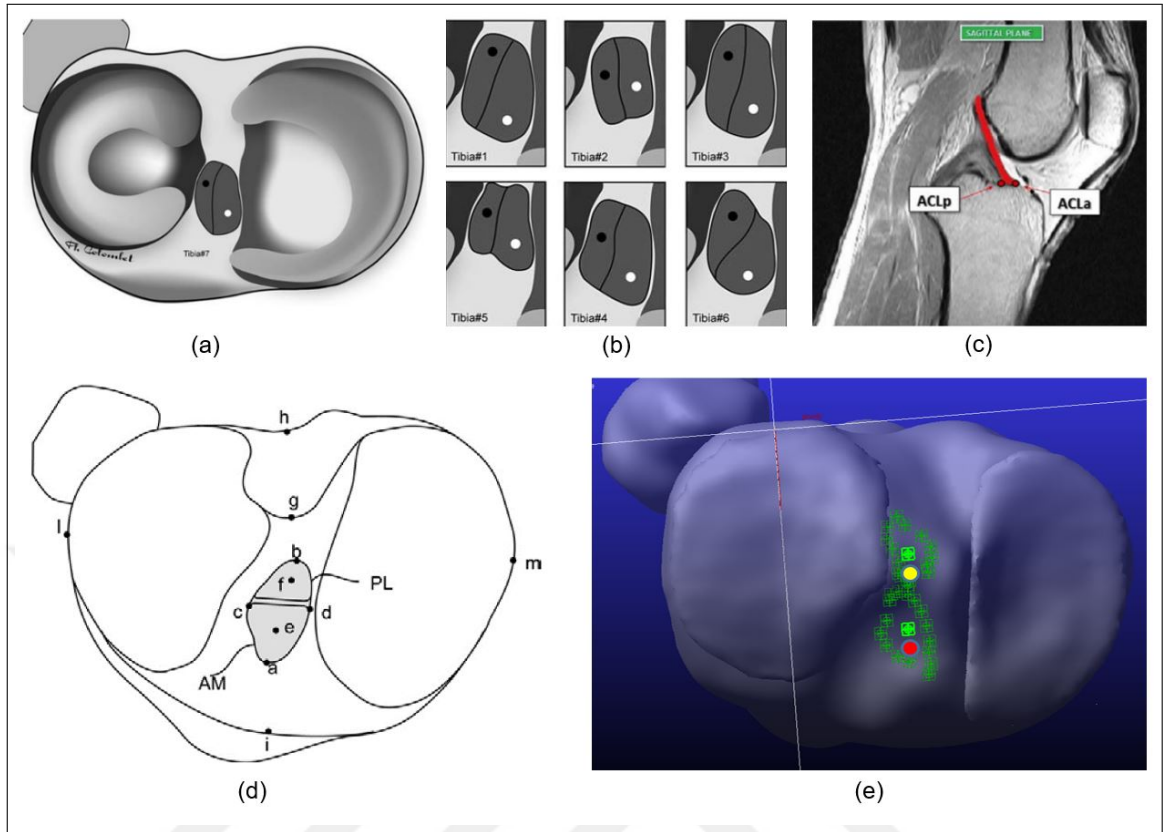


**Figure 2.5** Anterior Cruciate Ligament (ACL) attachment sites of proximal tibia head. (a) The central fibers of the anterior bundle of ACL (aACL) depicted as a white dot and posterior bundle of ACL (pACL) as a black dot [8]. (b) The appearance of remained six specimen samples [8]. (c) The localization of the anterior cruciate ligaments via magnetic resonance images [9]. (d) The illustration of anatomic landmarks and surgical sites for clarity of attachment locations [8]. (e) The location point cloud and calculated centroids selected for insertion on tibia model, where red dot is aACL and yellow dot is pACL.

along its medial line. Additionally, the localization of the ACL was determined using Innocenti's method (Figure 2.5c, 2.6c and d), which employed MRI data [9].

To enhance clarity, an illustration of the anatomic landmarks and surgical attachment sites is provided (Figure 2.5d). The tibial landmarks include: (a) the anterior extent of the ACL attachment area, (b) the posterior extent, (c) the medial extent, (d) the lateral extent, (e) the projection of the center of the AM bundle, (f) the projection of the center of the PL bundle, (g) the retro-eminence ridge, (h) the posterior border of the tibial plateau, (i) the anterior border of the tibial plateau, (l) the lateral border, and (m) the medial border [8].

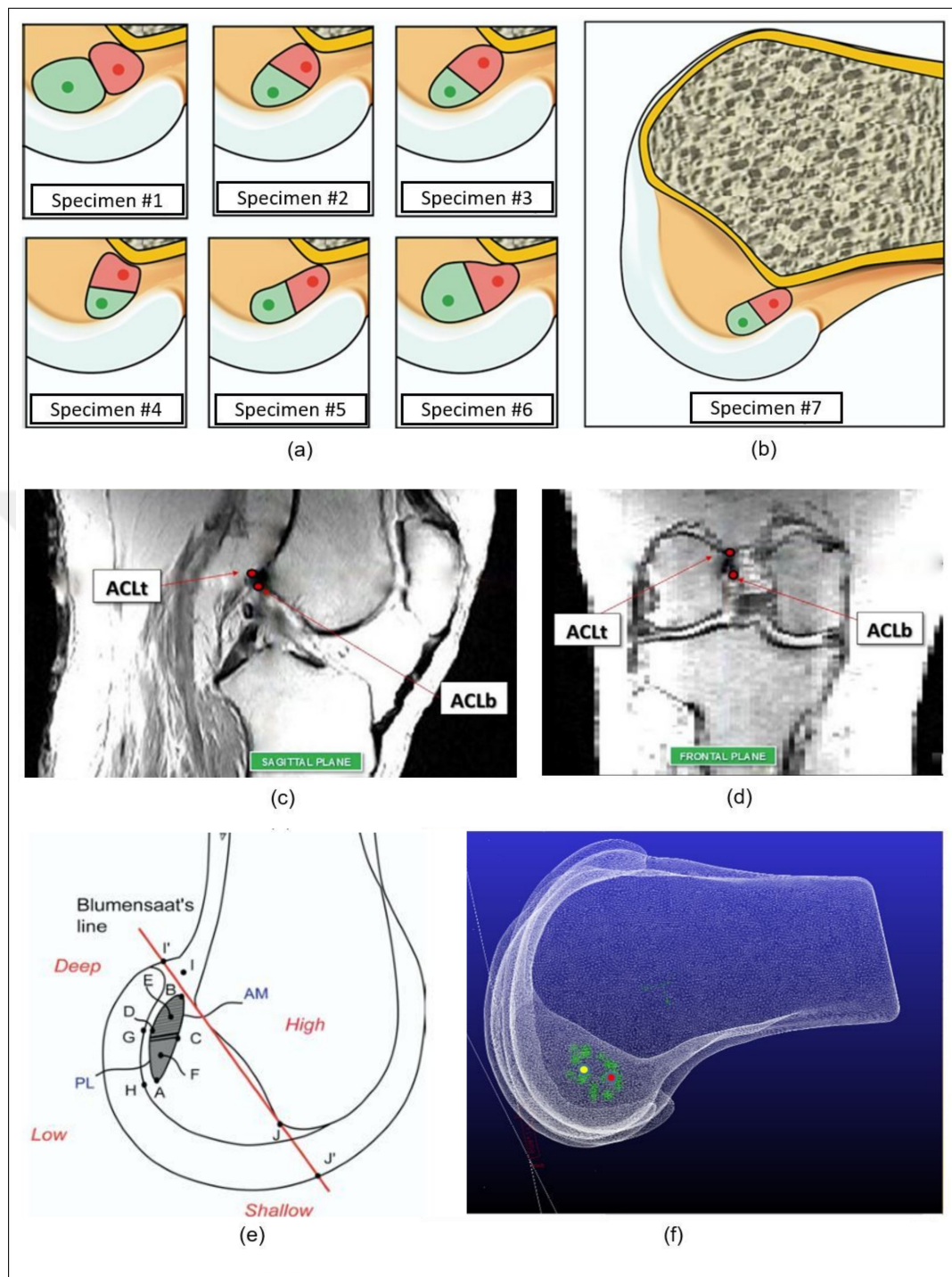


**Figure 2.6** The sagittal view of the right femur. (a) The projections of the femoral insertion sites (b) magnified view of the Antero-Medial bundle (AM) or anterior bundle of Anterior Cruciate Ligament (aACL) and the Postero-Lateral bundle (PL) or posterior bundle of the Posterior Cruciate Ligament (pPCL) insertions [8]. The localization of the posterior cruciate ligament insertion by means of magnetic resonance images (c) sagittal plane and (d) frontal plane[9], (e) diagram for illustration of surgical and anatomic landmarks [8], (f) the location point cloud and calculated centroids selected for insertion on femur model, where red dot is aACL and yellow dot is posterior fibers of Anterior Cruciate Ligament (pACL).

For the femur in the sagittal view, anatomic landmarks and navigation terminology in surgery are depicted (Figure 2.6e). These include: (A) the distal border of the ACL attachment area ("low" in the notch), (B) the proximal border ("high" in the notch), (C) the anterior border ("shallow" in the notch), (D) the posterior border ("deep" in the notch), (E) the projection of the center of the AM bundle, (F) the projection of the center of the PL bundle, (G) the posterior margin of the articular cartilage ("deep" in the notch), (H) the distal margin of the articular cartilage ("over-the-top" position or "high" in the notch), and (J) the most anterior point of the roof of the notch [8].

The point cloud sets provided by 'Natural knee' [5, 6] and calculated centroids highlighted, a red dot for the anterior bundle of anterior cruciate ligament (aACL) and a yellow dot for the posterior bundle of anterior cruciate ligament (aACL) (Figure 2.5 e and 2.6 f) were used as attachment points in knee model.

#### 2.4.2 Posterior Cruciate Ligament

The Posterior Cruciate Ligament (PCL) originates from the posterior tibia and travels anteromedially, passing posterior to the ACL, to attach to the lateral surface of the medial femoral condyle, as shown in Figures 2.7 and 2.8 [11]. Similar to the ACL, the PCL is considered a double-bundle structure, consisting of the Ante-Lolateral Bundle (ALB) and Postero-Medial Bundle (PMB). The PCL inserts onto the posterior intercondylar fossa, located between the tibial plateaus

The ALB, commonly referred to as the anterior bundle of the aPCL, attaches to the superolateral aspect of the fossa, as depicted in Figure 2.7a. Note that the specimen illustrated is a left tibia. The PMB, also known as the pPCL (Figure 2.7a), attaches to the inferomedial aspect of the fossa. A substantial portion of the posteromedial bundle extends below the posterior rim of the tibia [10].

The origin sites of the PCL bundles resemble a half-moon shape [11] on the

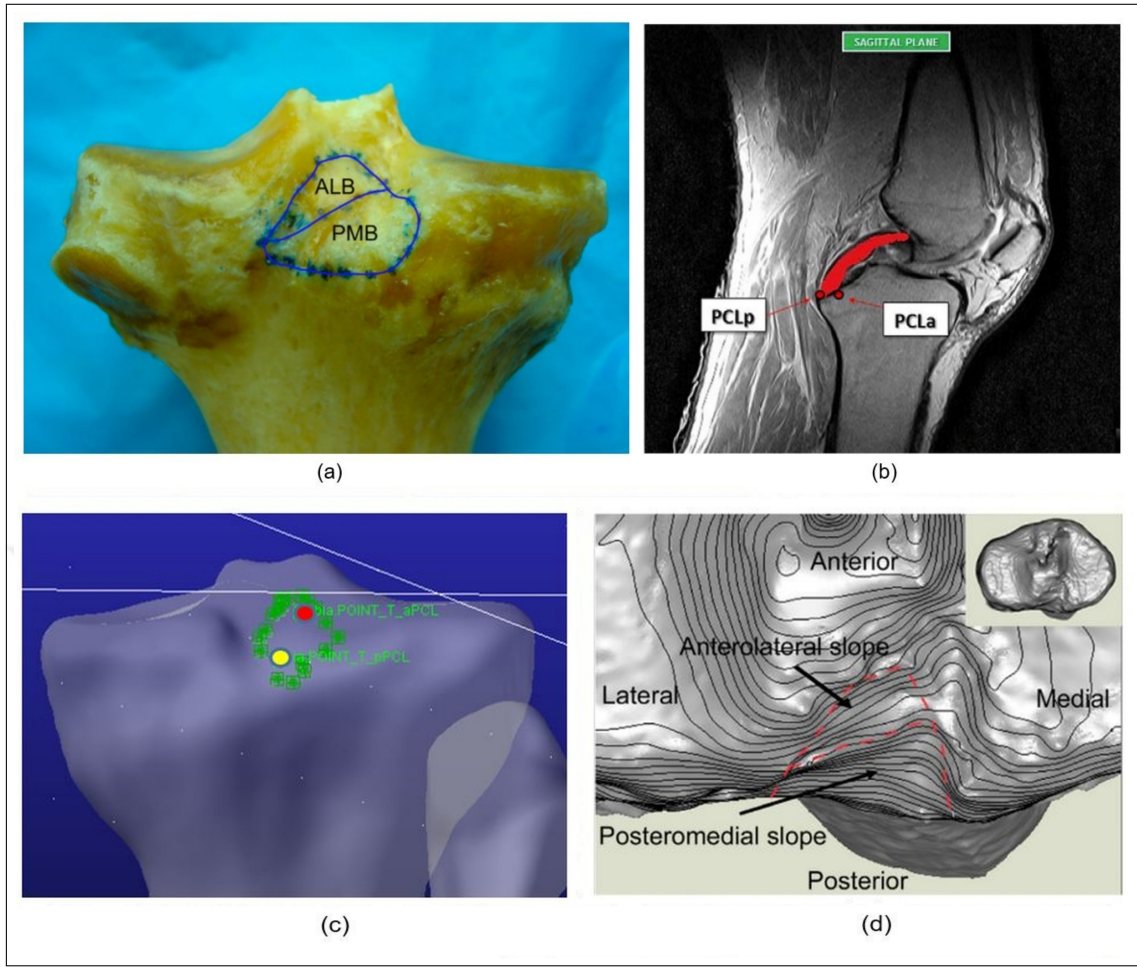


**Figure 2.7** (a) Posterior-proximal view of the left tibia showing the tibial insertion sites of the Antero-Lateral Bundle (ALB) or Anterior fibers of the Posterior Cruciate Ligament (aPCL) and the Postero-Medial Bundle (PMB) or posterior fibers of the Posterior Cruciate Ligament (pPCL) [10]. (b) The localization of the posterior cruciate ligament insertion on tibia [9]. (c) Posterior aspect of proximal tibia, illustrating the point cloud used to identify PCL bundles, where red dot is aPCL and yellow dot is pPCL. (d) The isohips image of the proximal head of the left tibia [10].

femur, as illustrated in Figure 2.8. These sites are located on the lateral surface of the medial femoral condyle (Figure 2.8c). Analysis of all three axes revealed that the mediolateral axis provided the most consistent reference for identifying the tibial insertion sites of the PCL (Figure 2.7b) and in MRI scans. Given its alignment along the sagittal plane and its anteroposterior tilt, investigations focused on sagittal slices [9].

Conversely, for the femoral insertion sites, both frontal and sagittal planes proved valuable insight (Figures 2.8a and 2.8b) due to the oblique orientation of the ligament attachment [9].

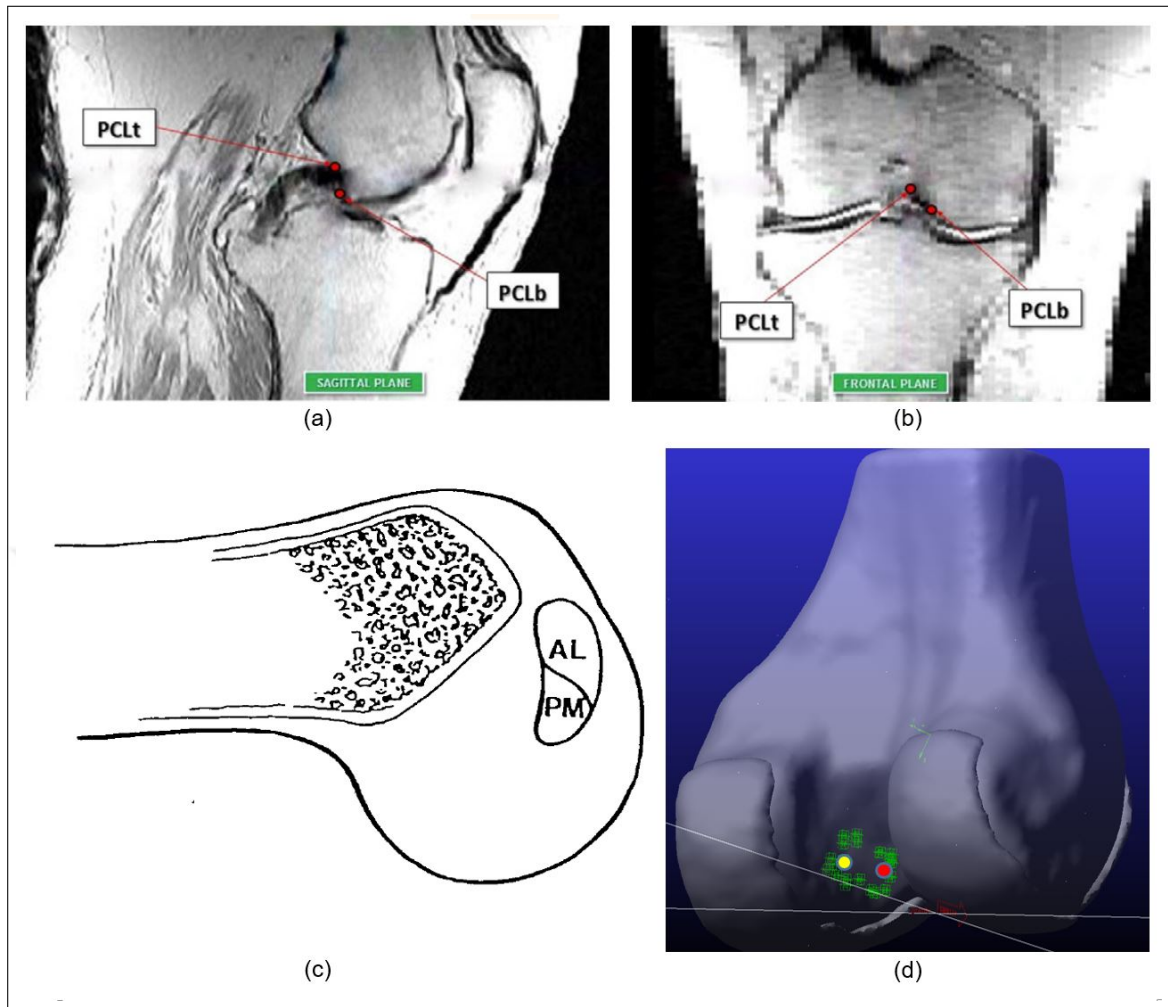


**Figure 2.8** The distal head of femur, the location of the femoral insertion points of the Anterior bundle of Posterior Cruciate Ligament (aPCL) on MRI (a) frontal plane (b) sagittal plane views [9]. (c) Antero-Lateral (AL) bundle or aPCL and the posteromedial (PM) bundle or Posterior fibers of the Posterior Cruciate Ligament (pPCL) origin sites are mapped on the inner walls of medial condyle [11]. (d) A point cloud and their centroid points on the medial condyle of the femur, where red dot represents attachment point of aPCL and yellow dot is posterior fibers of posterior cruciate ligament (pPCL).

The points identified on MRI and coordinate data obtained from the NK repository were combined with morphological studies available in the literature. For improved comprehension, the insertion sites of the ALB or aPCL, and the PMB or PCL, are illustrated on an isohips-mapped proximal tibial head [10]. A small inset image of the tibia in the upper right corner of Figure 2.7c indicates the orientation of the posterior intercondylar fossa.

This fossa was trapezoidal in shape, becoming wider inferiorly. The anterolateral bundle attached to the superolateral aspect of the fossa, referred to as the 'anterolateral slope,' while the posteromedial bundle attached to the anteromedial aspect of the fossa, termed the 'posteromedial slope' [10]. Centroids calculated from point cloud data provided in previous studies [5, 6], combined with anatomical and radiological studies, were used to define the attachment points for the mechanical knee model in ADAMS (Figures 2.7d and 2.8d).

### 2.4.3 Medial Collateral Ligaments

Most publications about the medial collateral ligament (MCL) focus on its anatomy and are conducted through the dissection of cadaveric human knees. The medial ligament complex of the knee comprises one large ligament and a series of capsular thickenings and tendinous attachments. This complex is commonly divided into two main bundles: the Superficial Medial Collateral Ligament (sMCL) and the Deep Medial Collateral Ligament (dMCL), also referred to as the tibial collateral ligament and the mid-third medial capsular ligament, respectively [12].

Some studies have further divided the dMCL into a proximal and a distal half [12, 14, 66]. The proximal half originates near the medial epicondyle of the femur and attaches to the medial meniscus, commonly referred to as the Menisco-Femoral Ligament (MFL). The distal half attaches to the medial meniscus and inserts into the proximal tibia, known as the Menisco-Tibial Ligament (MTL).

LaPrade et al. reported that the sMCL has one femoral and two tibial insertions [12]. Their investigation of eight cadaveric samples identified the Medial Epicondyle (ME) as the primary osseous anatomic landmark (Figure 2.9). They noted that the adductor tubercle is located 12.6 mm proximal and 8.3 mm posterior to the ME, while the gastrocnemius tubercle is 9.4 mm distal and 8.7 mm posterior to the adductor tubercle. Additionally, the gastrocnemius tubercle is positioned 6.0 mm (range: 4.4 - 8.9 mm) proximal and 13.7 mm posterior to the ME. The femoral origin of the

sMCL was located, on average, 3.2 mm proximal and 4.8 mm posterior to the ME. Furthermore, no firm attachment between the sMCL and dMCL was observed [12].

Liu et al. conducted a morphology study of the MCL and found that the anterior part of the sMCL coursed vertically, while its posterior part was oblique [14]. These findings align with previous modeling studies by Crowninshield et al., Blankevoort et al., and Abdel-Rahman et al. [67, 68, 69, 46, 48]. Crowninshield’s study categorized the medial collateral ligament into an Anterior Bundle of the Medial Collateral Ligament (aMCL), a Posterior Bundle (pMCL), an Oblique Bundle (oMCL), and a Deep Bundle (dMCL) [67].

Blankevoort’s study further divided the MCL into five parts, three corresponding to the sMCL—namely, the anterior fibers (aMCL), inferior fibers (iMCL), and posterior fibers (pMCL)—and two corresponding to the dMCL, consisting of an anterior bundle and a posterior bundle [68, 69]. Abdel-Rahman’s two modeling studies provided additional insight: the first model, conducted in two-dimensional space, included four fibers (aMCL, pMCL, oMCL, and dMCL) [46], while the subsequent three-dimensional model simplified the MCL to three fibers (aMCL, oMCL, and dMCL), omitting the pMCL [48].

Morphological and modeling studies consistently highlight the dMCL as a distinct structure. The overlying sMCL, which lacks strong interconnection with the dMCL, is frequently divided into anterior (aMCL), posterior (pMCL), and oblique (oMCL) bundles. However, there is ongoing debate about whether the Posterior Oblique Ligament (POL) is a separate structure or merely a fiber bundle of the sMCL [12, 13].

#### **2.4.4 Superficial Medial Collateral Ligament**

The Superficial Medial Collateral Ligament (sMCL), also known as the Tibial Collateral Ligament (TCL), has quantitatively defined insertion sites relative to osseous landmarks in dissected knees. LaPrade et al. reported that the adductor tubercle is

located 12.6 mm (range: 9.0 - 15.2 mm) proximal and 8.3 mm (range: 5.9 - 11.6 mm) posterior to the medial epicondyle. Additionally, the gastrocnemius tubercle was found to be 9.4 mm (range: 7.1 - 11.8 mm) distal and 8.7 mm (range: 6.8 - 12.5 mm) posterior to the adductor tubercle, as well as 6.0 mm (range: 4.4 - 8.9 mm) proximal and 13.7 mm (range: 10.8 - 15.8 mm) posterior to the medial epicondyle [12, 13].

The study conducted by Liu et al. identified that the femoral insertion of the sMCL spans the medial femoral epicondyle. Using a 3D digitizing probe, they measured distances between the centroids of ligament insertions and osseous landmarks. The distances from the centroid of the femoral insertion of the sMCL to the Medial Femoral Epicondyle (MFE), Adductor Tubercle (AT), and Medial Gastrocnemius Tubercle (MGT) were reported as  $2.9 \pm 0.8$  mm,  $16.5 \pm 1.6$  mm, and  $14.7 \pm 4.5$  mm, respectively. Illustrations and diagrams of the medial aspect of the knee were provided to depict osseous landmarks, insertion sites, and the possible span of the medial ligament (Figure 2.9a, b, and c) [12, 13, 14].

The distal attachment of the sMCL, corresponding to its tibial insertion, was described as oval-shaped and elongated (Figure 2.9a and c), located just anterior to the medial crest of the tibia (Figure 2.9d) [12, 13]. According to quantitative measurements by Liu et al., the average distance between the centroid of the sMCL's tibial insertion and the tibial joint line—defined as the edge of the articular cartilage surface of the medial tibial condyle—was  $62.4 \pm 5.5$  mm (range: 54.7 - 71.5). The mean overall length of the sMCL, measured from the centroid of its femoral insertion to the centroid of its distal tibial insertion, was  $100.7 \pm 9.5$  mm (range: 90.0 - 117.1) [14].

Point cloud datasets (Figure 2.9e) from the NK study [5, 6] were utilized to define the ligament's attachment points. Selected point clouds (Figure 2.9f), with red dots representing femoral centroids and yellow dots representing tibial centroids, were calculated as the insertions of the sMCL (Figure 2.9g). These points were subsequently applied as attachment sites in the knee model.

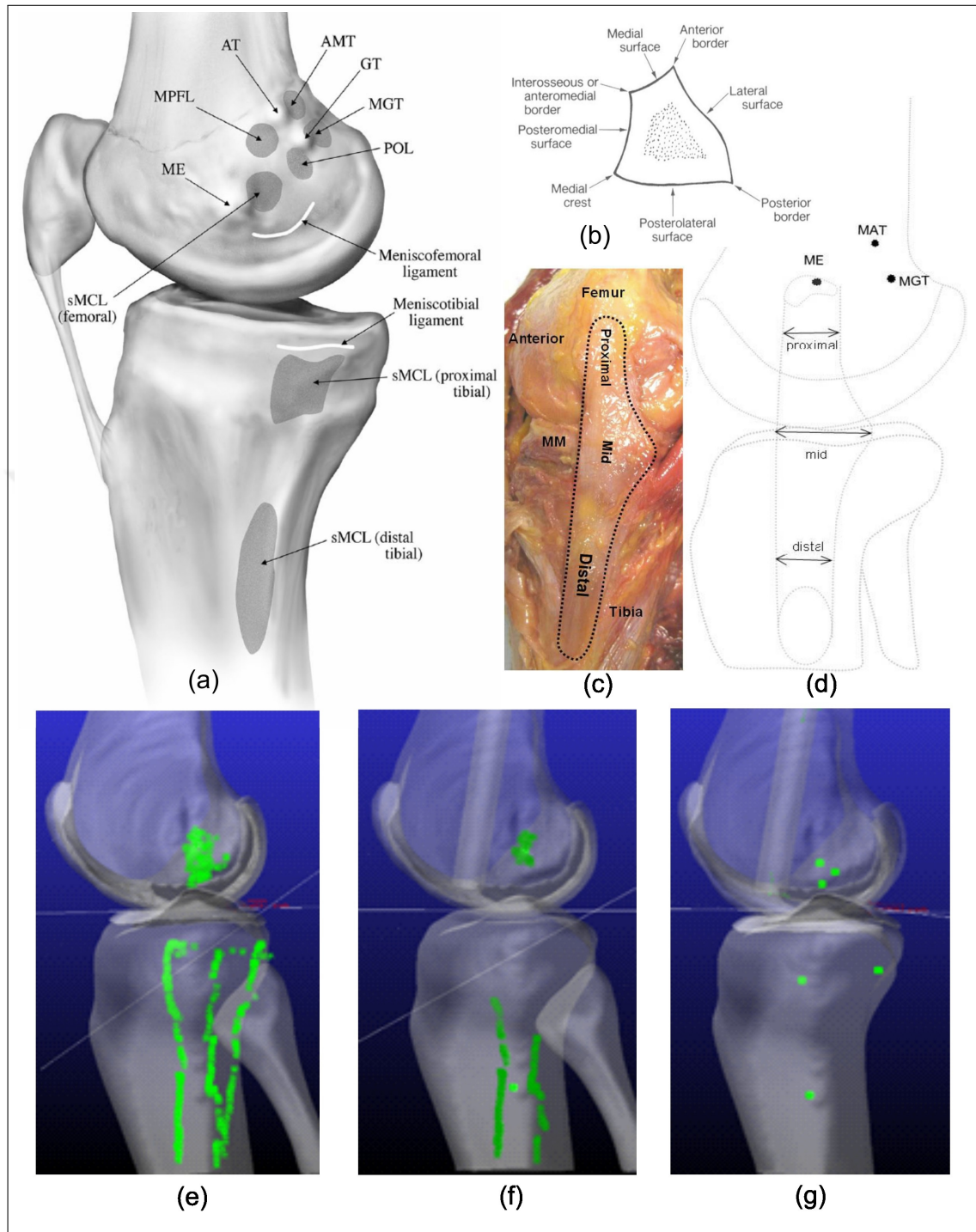


**Figure 2.9** Medial aspect view of right knee. (a) Illustration of the bone landmarks and insertion sites of ligamentous structures [12, 13]. (b) Dashed region shows medial collateral ligament on the photo of dissected cadaveric knee. (c) Diagram of medial collateral ligament were shown schematically [14]. (d) The cross section of tibia bone for qualitative allocating attachment site borders. (e) The point cloud measured for medial ligamentous attachment site, (f) the selected cluster and (g) the centroids (red dot) for insertion point of superficial Medial Collateral Ligament (sMCL).

### 2.4.5 Deep Medial Collateral Ligament

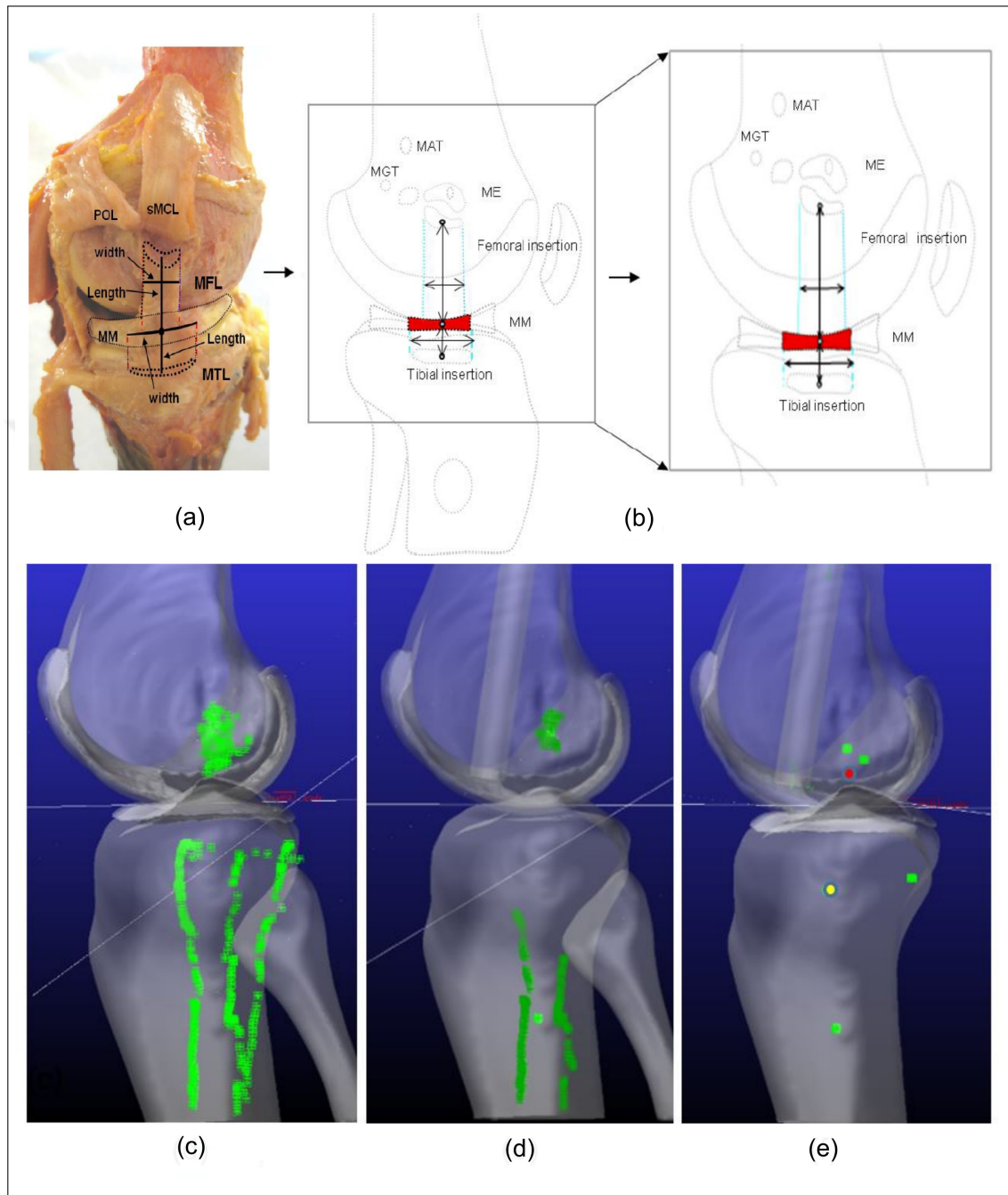


**Figure 2.10** Medial aspect view of left knee. (a) Superficial Medial Collateral Ligament (sMCL) reflected superiorly and posterior oblique ligament reflected postero-proximal to show deep Medial Collateral Ligament (dMCL) dashed region on dissected cadaveric left knee. (b) Simplified illustration of the bone landmarks and insertion sites of dMCL [14]. (c) The point cloud measured for medial ligamentous attachment site, (d) the selected cluster and (e) the centroids (red dot) for insertion point of dMCL.

The deep medial collateral ligament (dMCL) which also known as Mid-third medial capsular ligament, comprised menisiofemoral and meniscotibial ligament compo-

nents (Figure 2.10a and b). The meniscomfemoral part was almost three times longer than meniscotibial part [14]. Distally, the dMCL is attached to the edge of the articular cartilage of the medial tibial plateau. Proximally it originates at slightly distal and deep to the femoral attachment of the sMCL (Figure 2.10a and b). This knowledge was useful for the comprehension of anatomy quantitatively, thanks to the research team [5, 6], there was available data for the specific specimen. The whole point cloud for medial ligamentous attachments on the femur and tibia (Figure 2.10c) was neatly separated for the dMCL (Figure 2.10d) and centroids were obtained red dots (Figure 2.10e) by averaging selected coordinate points.

#### 2.4.6 Oblique Medial Collateral Ligament

In earlier studies on the medial knee, the superficial medial collateral ligament (sMCL) was reported to have an oblique posterior portion, now recognized as the posterior oblique ligament (POL) [13]. Originating from the medial femoral condyle, the POL fans out distally and blends with the semimembranosus tendon (Figure 2.11a, and b). The femoral attachment of the POL is located proximal and posterior to the femoral attachment of the sMCL (Figure 2.11a and b). Its mean centroid is positioned 7.7mm distal and 6.4mm posterior to the adductor tubercle (AT) and 1.4mm distal and 2.9mm anterior to the Gastrocnemius Tubercle (GT) [13].

The distal attachment of the POL primarily anchors to the posteromedial aspect of the medial meniscus, the meniscotibial portion of the posteromedial capsule (Figure 2.11a and b), and the posteromedial region of the tibia [13, 15]. Point cloud datasets from the "Natural Knee" study (Figure 2.11c) [5, 6] were utilized to define the attachment points of the oblique medial collateral ligament (oMCL). Selected points are depicted as green dots (Figure 2.11d), with red dots marking femoral centroids and yellow dots marking tibial centroids, calculated as the insertions of the oMCL (Figure 2.11e). These centroids were subsequently incorporated as attachment sites in the knee model.

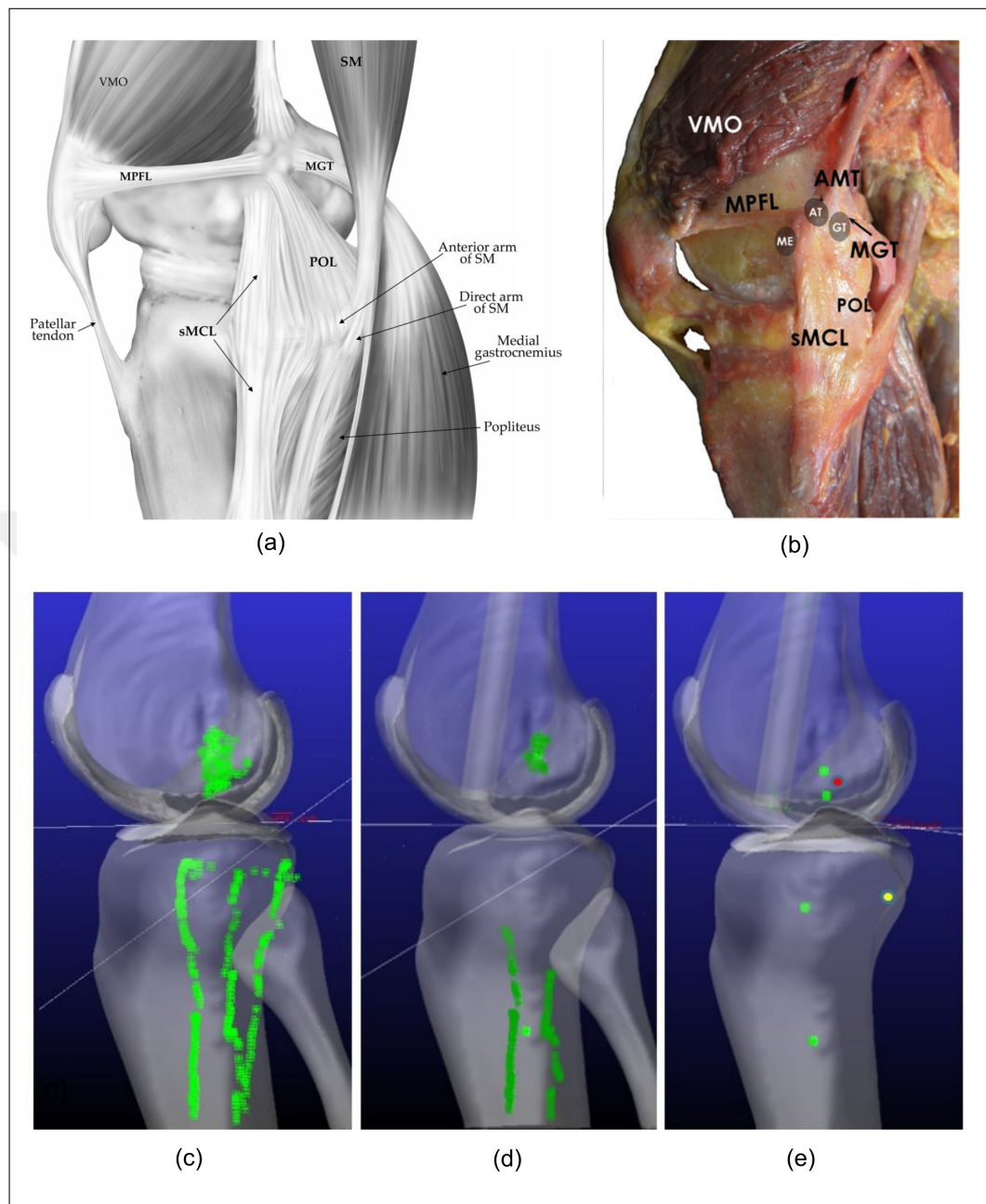


**Figure 2.11** Medial aspect view of left knee. (a) The illustration and (b) The specimen photo of the posterior oblique ligament (POL/oMCL) fanned out inferiorly and located at the posterior of sMCL [13]. (c) The relationship of the semimembranosus capsular expansion, the posterior oblique ligament (POL) or oblique fibers of Medial Collateral Ligament (oMCL), and the posteromedial meniscus [15]. (d) The point cloud measured for medial ligamentous attachment site, (e) the selected cluster and (f) the centroids (red and yellow dot) for insertion point of POL or dMCL.

#### 2.4.7 Lateral Collateral Ligament

The Fibula bone does not involve the knee joint directly. The fibula head provides an insertion site for both the static stabilizers (ligaments) to resist posterior

translation, varus, and external rotation forces and the dynamic stabilizers (biceps femoris tendon) to resist anterolateral-antromedial rotary instability of the knee joint [16, 57]. The main ligament at the lateral side of the knee is the Lateral Collateral Ligament (LCL) which is also widely known as the also known as the Fibular Collateral Ligament (FCL) [57, 13].

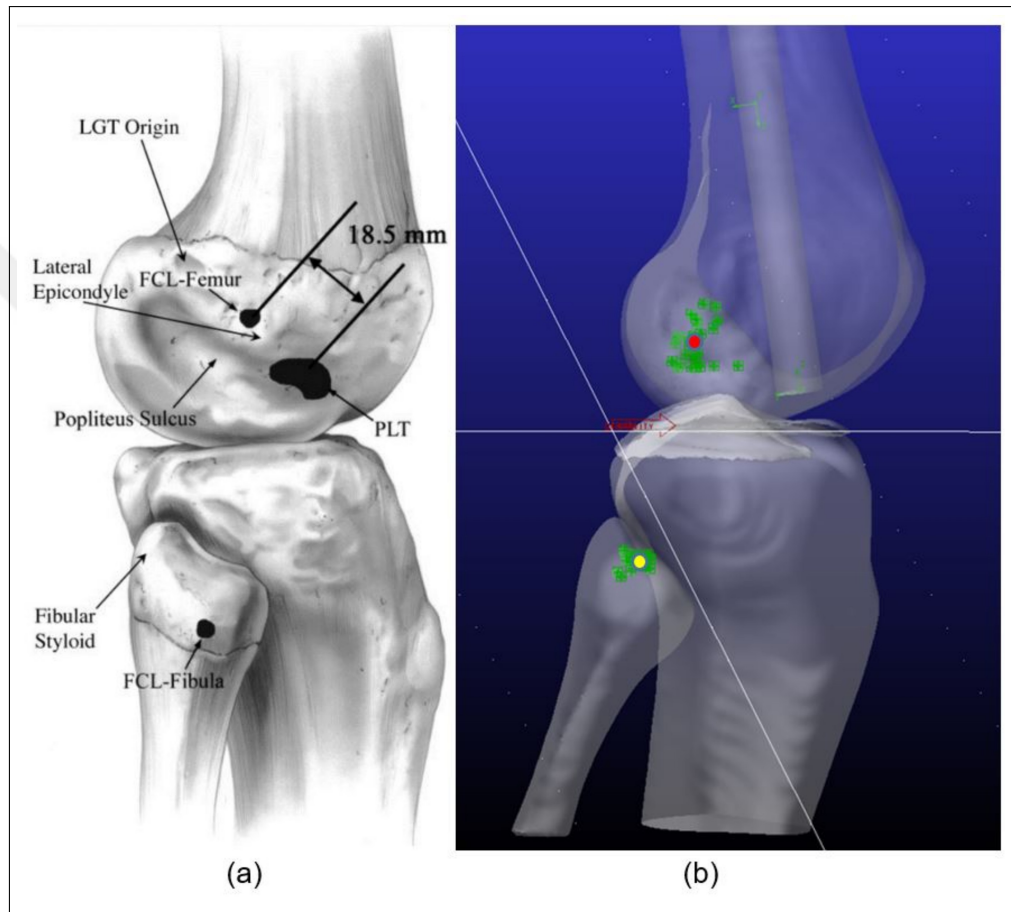


**Figure 2.12** Lateral view of the right knee. (a) The black regions show the insertion area of the Lateral Collateral Ligament (LCL) or Fibular Collateral Ligament (FCL) [16]. (b) The insertion points are calculated centroids of point clouds obtained from cadaver knee [5, 6] on femur red dot and on tibia yellow dot.

The main femoral attachment of the fibular collateral ligament is located in a small bony depression just posterior to the lateral epicondyle as shown in Figure 2.12a. Abbreviations in Figure stand for Lateral Gastrocnemius Tendon (LGT), Popliteus Tendon (PLT), and Fibular Collateral Ligament (FCL). The LCL coursed distally and attached to the lateral aspect of the fibular styloid process. The mean insertion point was 8.2 mm (range, 6.8 to 9.7) posterior to the anterior margin of the fibular head and

28.4 mm (range, 25.1 to 30.6) distal to the tip of the fibular apex [16]. The average attachment on the femur was found slightly proximal (1.4 mm; range, 0.8 to 2.7) and posterior (3.1 mm; range, 2.3 to 4.4) to the lateral epicondyle. The mean length of the LCL was found 69.6 mm (range 62.6 to 73.59) by LaPrade et al. [16] and  $61.0 \pm 4.7$  mm by Ishigooka et al. [56]. The point cloud sets provided by NK [5, 6], green dots were selected point clouds, red dots for the femur, and yellow dots for the tibia calculated centroids as insertions of the LCL (Figure 2.12a) were used as attachment points in the knee model. Centroids calculated from point cloud provided in the literature [5, 6] combined with anatomy and radiology studies attachment point were defined for the mechanical model of the knee in ADAMS (Figure 2.12b).

#### 2.4.8 Oblique and Arcuate Popliteal Ligaments

The Oblique Popliteal Ligament (OPL) is the largest structure on the posterior aspect of the knee and is named because of its shape. According to the findings by Wu et. al., the OPL originated from primarily the posterior surface of the posteromedial tibial plateau, and fibers from the posteromedial part of capsule blended to it as a secondary medial origin [60]. Both origins blended and coursed obliquely to the superior and lateral direction and attached to the fabella or the tendon of the lateral head of the gastrocnemius merging to the posterior capsule over the lateral femoral condyle. It was consistent with the anatomy study conveyed by LaPrade et al (Figure 2.13a). They reported lateral expansion of the semimembranosus tendon was the component of the OPL.

On the proximal lateral insertion site, OPL was found attached to the sesamoid bone or cartilaginous fabella (Figure 2.13a), the meniscomfemoral portion of the posterolateral joint capsule, and a plantaris muscle on their all specimens [13]. Where Semi-Membranosus (SM), sMCL, POL, PCL, OPL, and FCL were illustrated in Figure for clarity of the attachment site. The length and width of the OPL were measured at an average of 48 mm and 9.5 mm wide at its medial origin and 16.4 mm wide at its lateral attachments. According to reviewed reports [13, 60, 59], they were in consensus

about the proximal lateral insertion site merged in the fabella if it exists or blended with posterolateral joint capsule.

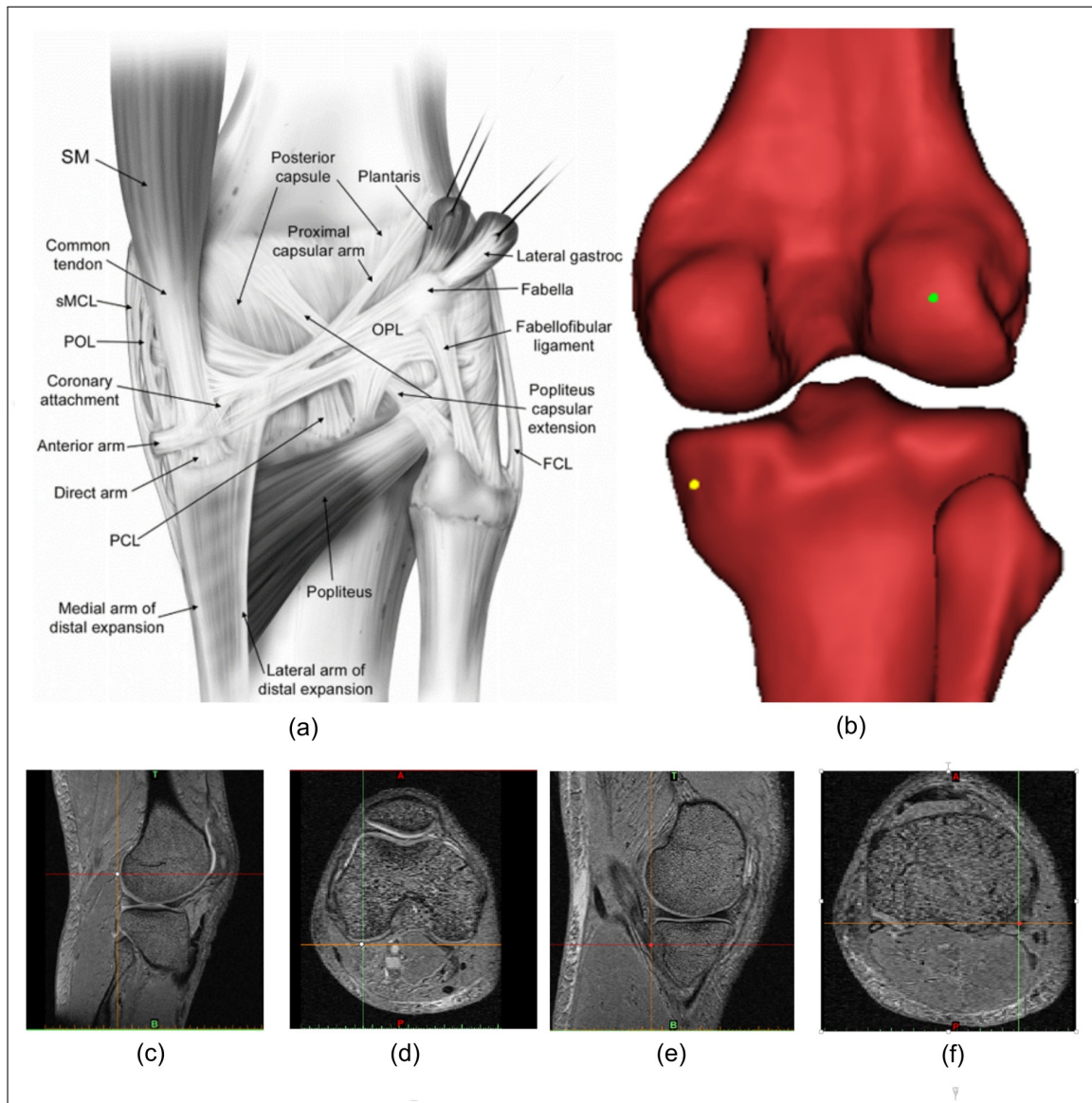


**Figure 2.13** Posterior view of the right knee. (a) The posterior knee fibrous complex, Oblique Popliteal Ligament (OPL), POL, PCL, sMCL and FCL/LCL [16]. (b) The green and yellow dot attachment locations of OPL on femur and tibia. The magnetic resonance images (c) sagittal plane and (d) axial plane view of selected point for the attachment of popliteal ligament on femur. MRI for locating OPL on tibia (e) sagittal and (f) axial slices.

The Arcuate Popliteal Ligament (APL) strengthens the joint capsule posterolaterally. It arises from the posterior aspect of the fibular head, passes superior and medially over the tendon of the popliteus, and spreads over the posterior surface of the knee joint. Its development appears to be inversely related to the presence and size of

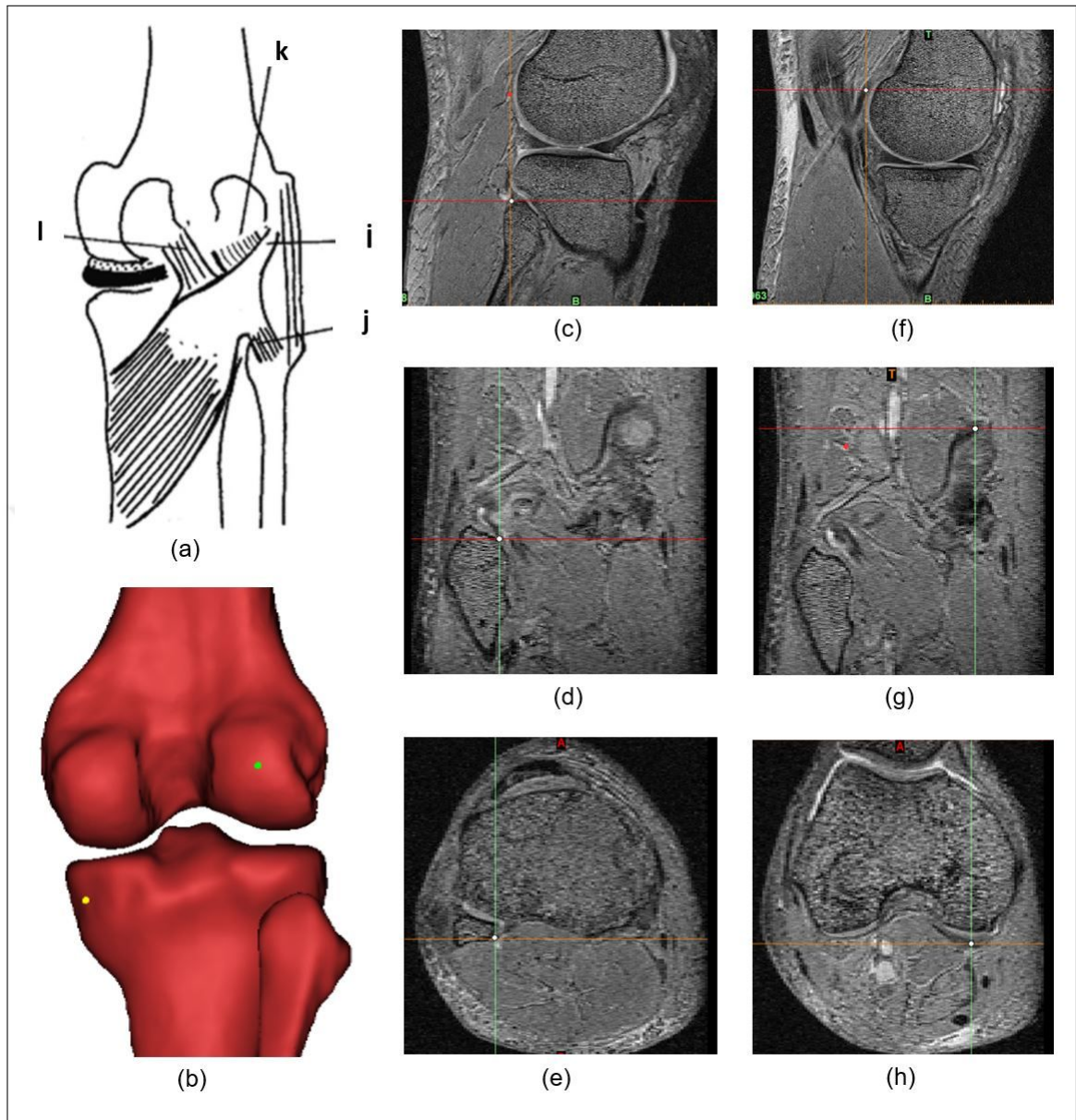


**Figure 2.14** Posterior view of the right knee. (a) The morphological study show Y shaped fibrous complex at posterior of the right knee [17]. (b) The green dot and yellow dot are insertions on femur and tibia respectively. The MRI slices for (c) sagittal, (d) frontal and (e) axial planes for localization of arcuate popliteal ligament on femur and (f) sagittal, (g) frontal and (h) axial planes for the localization of arcuate popliteal ligament on tibia.

a fabella (sesamoid bone at posterolateral part of the knee) in the proximal attachment of the lateral head of gastrocnemius [58]. Both structures were thought to contribute to the posterolateral stability of the knee. Ishigooka et al. investigated 78 (bilateral knee samples of 39 subjects) cadaver knees for Postero-Lateral Structure (PLS) of the knee [56]. They separated structure into Popliteus Tendon (PT), LCL, Arcuate Ligament

(AL), Fabella Fibular Ligament (FFL) and Popliteo-Fibular Ligament (PFL). PFL exists in all knees. Predominantly, 81.6 % of knees had both AL and FFL, 15.8 % of knees had only APL but just 2.6 % knees had FFL without AL [56]. The morphology study by Paraskevas et. al. reported that the fibrous band was found on 22 out of 24 specimens in PLS. It was a Y-shaped ligament (Figure 2.14a) known as the arcuate ligament. One arm of the Y is formed by the fibular origin and the other arm by the meniscal origin of the popliteus muscle [17]. Where letters denote (i) - main popliteal tendon, (j) - fibular origin, (k) - meniscal origin, and (l)-capsular origin of an arcuate popliteal ligament. To our knowledge, LaPrade et. al.'s reports did not discuss arcuate ligament (AL) or arcuate popliteal ligament (APL). However, in their anatomy study on the lateral aspect of the knee, it was seen that the band originated on the fibular head coursed superior medially, and blended into the posterior capsule (Figure 2.13a) and pointed with arrows as popliteus capsular extension [13]. It was also named as posterior capsule arcuate popliteal ligament by Abdel-Rahman et al.

#### **2.4.9 Posterior Capsule Lateral and Medial Structures**

The investigation on posterior capsule contribution to the knee joint was found to be significant [58, 70, 13, 15]. Both Medial fibers and Lateral fibers of the Posterior Capsule (mCAP and lCAP) insertion points coincide with the OPL and APL [67, 48]. Crowninshield et al. and Abdel-Rahman et al. conducted the three-dimensional knee model studies. There were a few models which were included posterior capsule fibers in their model. In both studies popliteal complex involving OPL and APL cross diagonally and vertices connected vertically by thickened capsular structure. The femoral attachment of mCAP was located at the same coordinate point as the femoral attachment of APL where tibia insertion was at the point of tibial insertion of OPL. Similarly, lCAP are defined at the femur attachment point of the OPL and inferiorly attached to the same point where the APL originated at the tibia.

#### **2.4.10 Antero Lateral Ligament (ALL)**

Anatomy investigation referred existence of connective tissues at the anterolateral site which is different structure than ilio-tibial tract. The passive resisting connective tissue on the lateral side of the knee have been known as the LCL. Although capsule confining synovial fluid surrounds joint, it is not distinguished as ligament or load carrying tissue in both anatomy and modeling studies. Anatomy study underwent by Claes et. al. in 2013 showed that lateral compartment of the knee joint has more passive resisting ligamentous structure than previously known [18]. They identified 40 antero-lateral ligament within 41 sample knees. Therefore it is reasonable to consider it in the present model.

Although the prevalence of the Antero Lateral Ligament (ALL) is high in previous studies, there are few of them resulted with low number of existence in sampled specimens and varies highly with race [61, 71]. In order to check the existence of the ALL which also known as the Antero-Lateral Structure (ALS), segmentation software is utilized. Since the resolution of MR images are not high enough and the ALL is fused into capsule by the nature, it is difficult to segment from fibrous structure. However, the bundle like segment can be extracted within the fibrous structure of capsule as shown in Figure 2.15b. It origins at the femoral epicondyle traverse obliquely to the anterior lateral site of tibia. Insertion sites are mapped according to previous anatomy studies [18, 61, 71] combined with segments cultivated by software.

### **2.5 Computational model (CM)**

#### **2.5.1 Ligament placement and tension modelling**

Figure 2.16 illustrates the attachment locations and line of action for all tensile elements corresponding to ligaments and capsules. A total of fifteen tensile elements are defined. The origin and insertion points are established with respect to coordinate

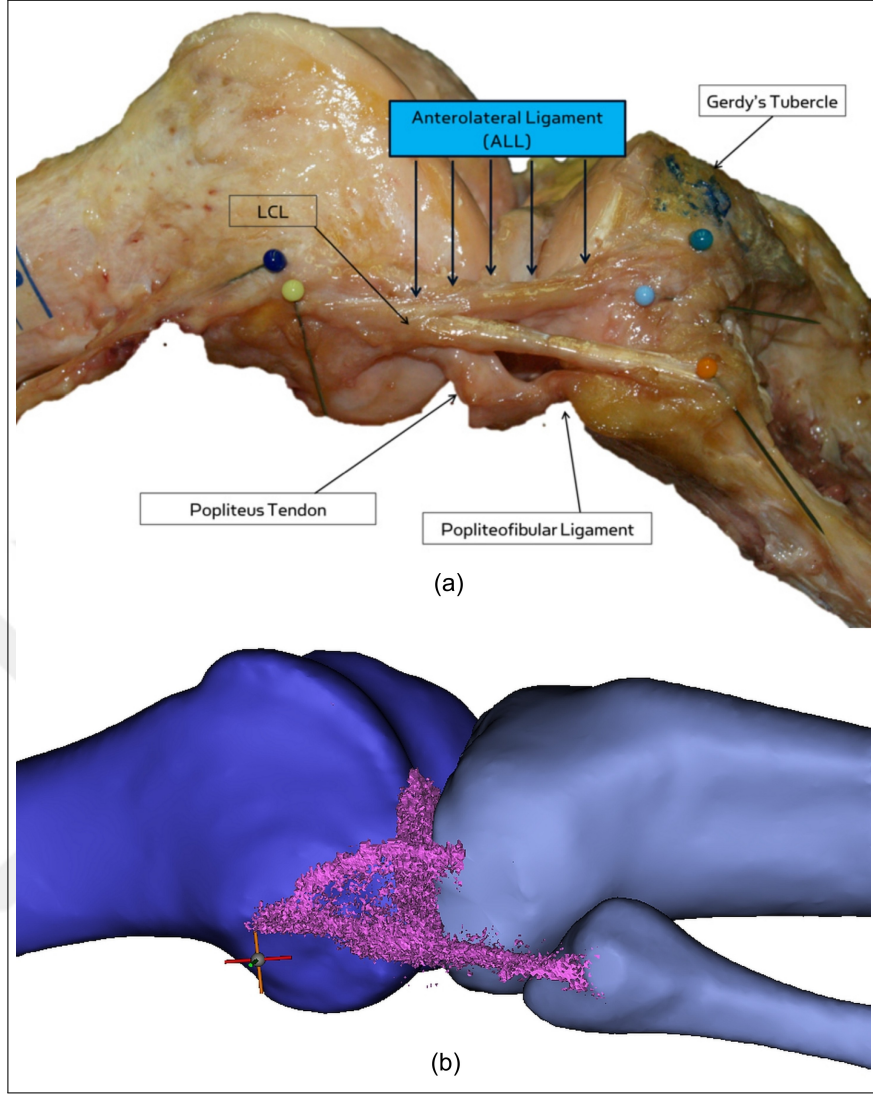


**Figure 2.15** Antero-Lateral Ligament (ALL): (a) Representation on a dissected knee, as described by Claes et al. [18], and (b) its anatomical structure identified above the lateral collateral ligament through segmentation of MRI data from the specimen.

points introduced in Section 2.4. Following the footsteps of earlier studies, cruciate ligaments are constructed as double bundles: an anterior bundle of the Anterior Cruciate Ligament ( $ACL$ ) denoted as  $aACL(a_t, a_f)$  and a posterior bundle denoted as  $pACL(b_t, b_f)$  [8]; similarly for the Posterior Cruciate Ligament ( $PCL$ ) an anterior bundle denoted as  $aPCL(c_t, c_f)$  and a posterior bundle denoted as  $pPCL(d_t, d_f)$  [11, 10].

The oblique and deep Medial Collateral Ligaments ( $oMCL$  and  $dMCL$ ) in the medial compartment of the knee are also presented as pairs of force elements due to their wide nature. The  $oMCL$  has anterior bundle  $oMCLa(g_t, g_f)$  and posterior

**Table 2.1**

The locations of attachment point of ligamentous line acting component with respect to global coordinate system.

|              | <b>Femoral attachments</b> |          |         | <b>Tibial attachments</b> |          |          |
|--------------|----------------------------|----------|---------|---------------------------|----------|----------|
| <b>Name</b>  | X (mm)                     | Y (mm)   | Z (mm)  | x (mm)                    | y (mm)   | z (mm)   |
| <b>aACL</b>  | 17.0001                    | -16.2146 | 22.5557 | 19.6239                   | -34.8145 | 1.3771   |
| <b>pACL</b>  | 16.2198                    | -19.1012 | 16.5694 | 20.4347                   | -22.0393 | 4.1704   |
| <b>aPCL</b>  | 26.3209                    | -30.7384 | 12.2267 | 20.2675                   | -9.3556  | -3.9126  |
| <b>pPCL</b>  | 32.4952                    | -25.4504 | 13.1972 | 24.1521                   | -6.0767  | -11.0674 |
| <b>sMCL</b>  | 63.1323                    | -27.8590 | 16.0696 | 37.7101                   | -28.7758 | -60.7743 |
| <b>oMCLa</b> | 62.5395                    | -23.1543 | 13.1531 | 62.5395                   | -8.9612  | -19.2325 |
| <b>oMCLp</b> | 62.5395                    | -23.1543 | 13.1531 | 62.5395                   | -8.9612  | -19.2325 |
| <b>dMCLa</b> | 61.3694                    | -26.9413 | 9.0653  | 61.3695                   | -32.2024 | -22.7930 |
| <b>dMCLp</b> | 61.3694                    | -26.9413 | 9.0653  | 61.3695                   | -32.2024 | -22.7930 |
| <b>LCL</b>   | -15.7148                   | -16.2141 | 20.8142 | -20.4867                  | -3.8983  | -24.9972 |
| <b>ALL</b>   | -15.7148                   | -16.2141 | 20.8142 | -20.4867                  | -3.8983  | -24.9972 |
| <b>mCAP</b>  | 50.8583                    | -4.3182  | 26.7505 | 52.3470                   | -9.0267  | -19.2761 |
| <b>lCAP</b>  | 0.6171                     | 1.7006   | 21.3141 | -1.4872                   | 1.1198   | -21.5357 |
| <b>OPL</b>   | 0.6171                     | 1.7006   | 21.3141 | 52.3470                   | -9.0267  | -19.2761 |
| <b>APL</b>   | 50.8583                    | -4.3182  | 26.7505 | -1.4872                   | 1.1198   | -21.5357 |

bundle  $oMCLp(h_t, h_f)$ , similarly, the  $dMCL$  has anterior bundle  $dMCLa(e_t, e_f)$  and posterior bundle  $dMCLp(f_t, f_f)$  [15, 12, 14]. The remaining ligaments, including the superficial Medial Collateral Ligament ( $sMCL(j_t, j_f)$ ), Lateral Collateral Ligament ( $LCL(l_t, l_f)$ ), Anterior Lateral Ligament ( $ALL(k_t, k_f)$ ), Oblique Popliteal Ligament ( $OPL(m_t, n_f)$ ) and Arcuate Popliteal Ligament ( $APL(n_t, m_f)$ ), are represented as single force elements, with attachments determined using MR images and anatomy studies [12, 14, 16, 55, 13, 56, 57, 17, 58, 59, 60, 18, 61]. Forces associated with the knee capsule, medial Capsule ( $mCAP(m_t, m_f)$ ), and lateral Capsule ( $lCAP(n_t, nl_f)$ ) are included as described in previous studies [67, 48, 15, 13, 70, 58].

The magnitudes of the ligament and capsule tensions are calculated using the force-elongation relationship as in Eq. 2.1 [20, 45, 48, 49, 50, 51, 69]. The force acting on the insertion points (Figure 2.16) along the line of sight carries tensile load only

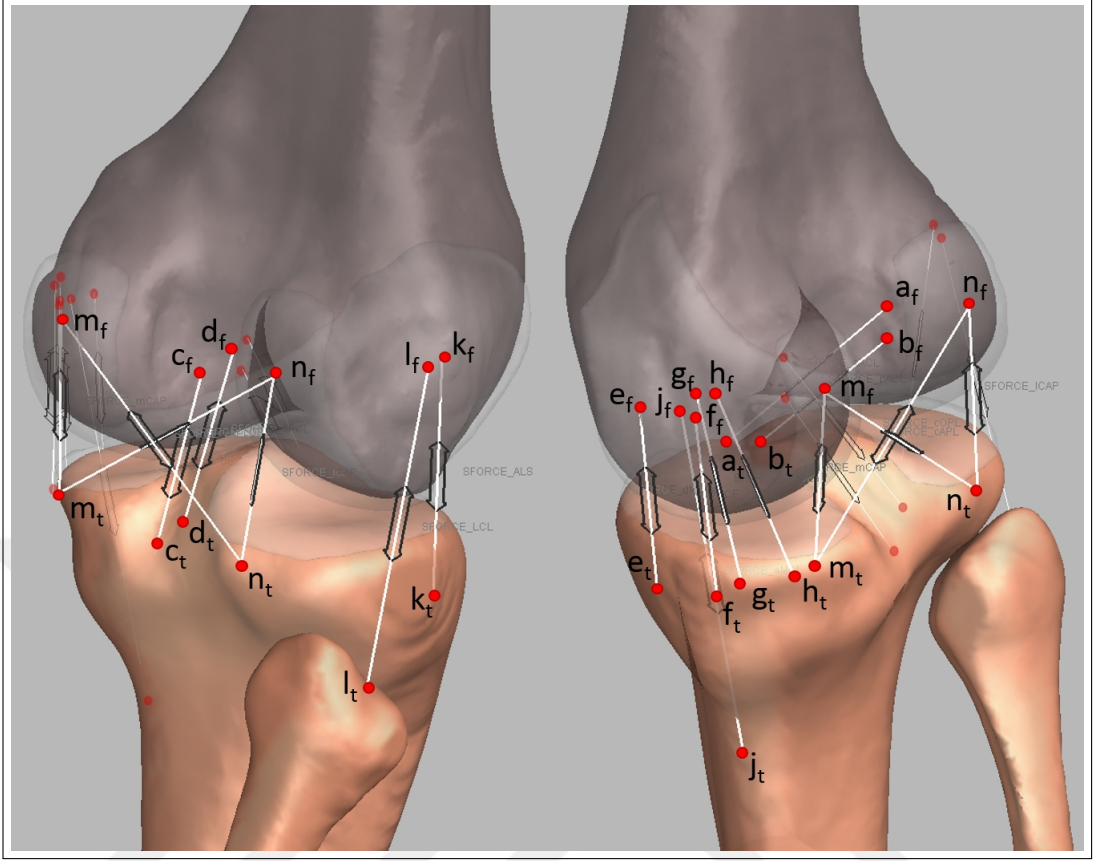


**Figure 2.16** Ligament insertion points are located using MRI and previous studies. Force element lines represent ligaments. The generated force on each tensile element are related to its elongation which is defined by the magnitude of current and reference vectors: aACL( $a_t, a_f$ ), pACL( $b_t, b_f$ ), aPCL( $c_t, c_f$ ), pPCL( $d_t, d_f$ ), dMCLa( $e_t, e_f$ ), dMCLp( $f_t, f_f$ ), oMCLa( $g_t, g_f$ ), oMCLp( $h_t, h_f$ ), sMCL( $j_t, j_f$ ), ALL( $k_t, k_f$ ), LCL( $l_t, l_f$ ), OPL( $m_t, n_f$ ), APL( $n_t, m_f$ ), mCAP( $m_t, m_f$ ) and lCAP( $l_t, l_f$ ).

when the  $i$ 'th ligament length  $L_i$  exceeds the unstrained slack length  $L_{0i}$ .

$$F_i = \begin{cases} 0 & \epsilon_i < 0 \\ K_{Qi}(L_i - L_{0i})^2 & 0 \leq \epsilon_i \leq \epsilon_{1i} \\ K_{Li}(L_i - L_{0i}(1 + \epsilon_1)) & \epsilon_{1i} < \epsilon_i \end{cases} \quad (2.1)$$

Due to intrinsic properties, ligamentous tension exhibits a 'toe region' in the initial stages of ligament strain, followed by a linear force-length relation. The former non-linear region has a quadratic relationship with the stiffness coefficient, denoted as  $K_{Qi}$ , while the latter is characterized by the stiffness parameter  $K_{Li}$ . The strain  $\epsilon_i$  in the ligament is calculated using the instantaneous length  $L_i$  and slack length  $L_{0i}$  as shown

in Eq. 2.2.

$$\epsilon_i = \frac{(L_i - L_{0i})}{L_{0i}} \quad (2.2)$$

Transition from toe to linear force region adjusted by  $\epsilon_1 i$ . Generic value of the constant is 0.03 in previous studies. However, we calculated  $\epsilon_1 i$  so the ltoe and linear force curve intersect on exact point. MATLAB code is developed for that purpose, see Appendix B. In Eq. (2.3), the slack length  $L_{0i}$  for each ligament is computed using the expansion constant  $e_i$  and the ligament length when the knee is in full-extension,  $L_{ei}$ .

$$L_{0i} = e_i L_{ei} \quad (2.3)$$

### 2.5.2 Contact surfaces and forces

Any two bodies contacting each other produce a contact force,  $F_c$  at the immersion site (Figure 2.17a).  $F_c$  can be decomposed into two orthogonal components, normal and parallel to the contact surface. The parallel force is neglected due to extremely low friction coefficient [72]. As a result  $F_c$  is reduced to:

$$F_c = \begin{cases} 0 & \delta < 0 \\ k \delta^p - \dot{\delta} C h(\delta, \delta_c) & 0 \leq \delta \end{cases}, \quad (2.4)$$

where  $\delta$  is the penetration depth as shown in Figure 2.17b and c,  $k$  is the contact stiffness coefficient of the immersion site,  $p$  is a coefficient defined by the materials' force deformation properties and  $C$  is the damping coefficient.  $h$  is a sigmoid-like step function with a unity gain and a transition interval  $\delta_c$ .

$$h = \begin{cases} 0 & \delta < 0 \\ 3((\delta - \delta_c)/\delta_c)^2 - 2((\delta - \delta_c)/\delta_c)^3 & 0 \leq \delta \leq \delta_c \\ 1 & \delta_c < \delta \end{cases}, \quad (2.5)$$

In this study,  $F_c$  is computed for cartilage-to-cartilage contact with  $k = 500$

N/mm,  $p = 1.5$ ,  $\delta_c = 0.01$  mm, and  $C = 5$  kg/s. [50, 73, 74, 75].

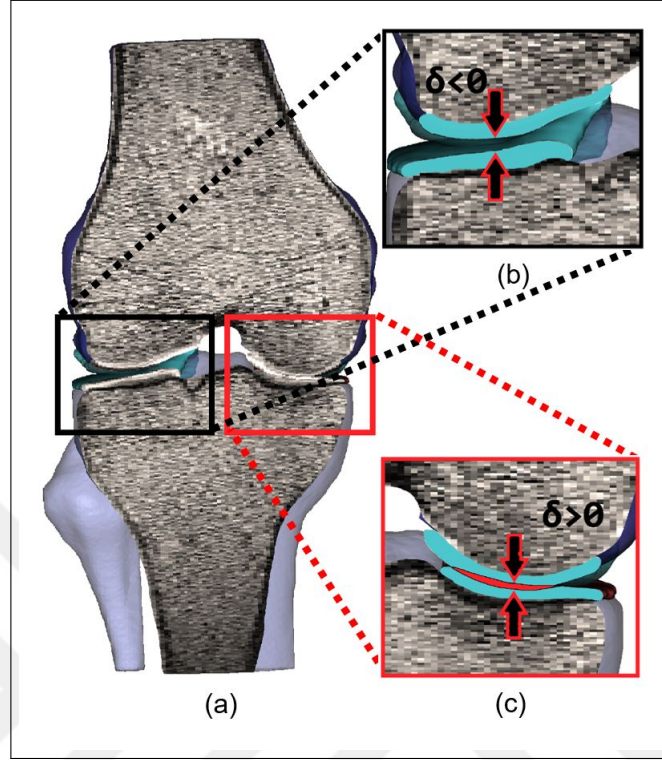


**Figure 2.17** Penetration depth ( $\delta$ ) for cartilage tissue is illustrated, (a) both possible condition of penetration are displayed for a knee joint, (b) when two cartilage surfaces do not penetrate ( $\delta > 0$ ), (c) when two cartilage surfaces penetrate ( $\delta < 0$ ).

We modeled the cartilages with a few millimeters of thickness based on MRI data. The cartilages of the distal condyles of the femur can immerse into the cartilages of proximal plateaus of the tibia confined by the material properties, which are on the order of sub-millimeters.

### 2.5.3 Initial condition and static equilibrium

In order to calculate force generated by a ligament as in Eq. (2.1), we need the slack lengths  $L_0$  for each ligament, as used in Eq. (2.2). The knee full-extension position is required for the calculation of the zero-load ligament lengths. However, since NK MR images were not acquired in full-extension, the resulting CM had to be brought to full-extension to estimate  $L_e$ , which is necessary to compute  $L_0$  for all the ligaments. After computing  $L_0$ , the model is relaxed from its full-extension posture to

an equilibrium state. At steady state, forces are balanced and the joint does not move, the model approaches to its starting NK posture.

#### 2.5.4 Kinematic validation

In this study we compared five motions of NK and CM joints during passive flexion motion. Anatomically, NK possesses all 6 degrees of freedom (DoF). Flexion-extension is exerted onto the knee joint externally, resulting in passive knee motion. During passive flexion, the knee joint continuously reorients and repositions, resulting in a coupling movement. The remaining knee rotations and translations become dependent on the flexion-extension motion [76]. Therefore, by exerting the same flexion-extension rotation (with the same start and end angles) on both NK and CM joints we expect the production of the same angular and linear trajectories. In other words, the remaining dependent motions of the 6 DoF, - abduction-adduction ( $\pi/2-\beta$ ), internal-external ( $\gamma$ ) rotation, medial-lateral ( $S_1$ ), anterior-posterior ( $S_2$ ), and inferior-superior ( $S_3$ ) translations (Figure 2.1) - are expected to be similar if CM is adequately constructed.

In order to validate the proposed multibody tibiofemoral model, the five dependent motions obtained from NK and computed from CM are compared quantitatively. In the NK model the tibia moves on a fixed femur as in Figure 2.18a. Since the femur was fixed to the experimental rig and the tibia was free to move as shown in Figure 2.2 [5, 6] the movement convention is named as 'Fixed Femur - Moving Tibia' (FF-MT). Consequently, the tibia was transformed from an initial position  $\mathbf{O}_{tib_i} = (X_{Ti}, Y_{Ti}, Z_{Ti})$  to a final point  $\mathbf{O}_{tib_f} = (X_{Tf}, Y_{Tf}, Z_{Tf})$  by three rotations and three translations.

For convenience, and to make our experiments compatible with the Clinical Joint System as described by Grood and Suntay, the flexion-extension movement was performed on the femur, constrained by a revolute joint hinged at the femoral origin and attached to the GCS as shown in Figure 2.18b [44]. Mechanically, the tibia is connected to the GCS with a perpendicular joint located at the tibial origin.

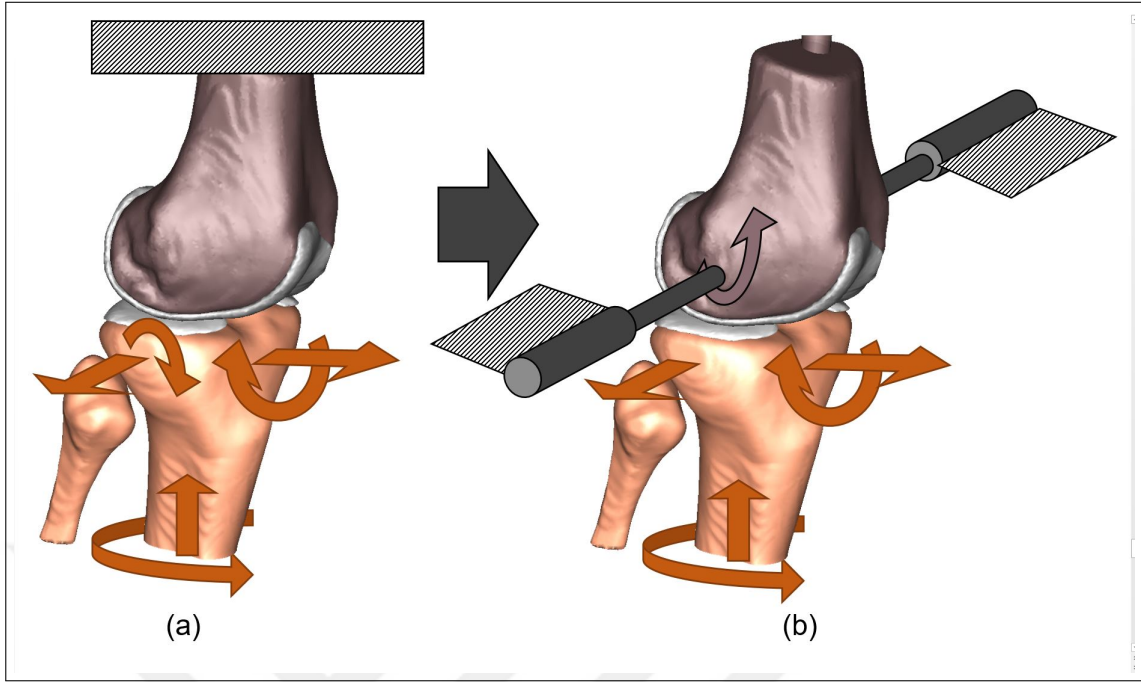


**Figure 2.18** (a) Tibia moving with respect to the femoral coordinate system which is fixed to the Global Coordinate System (GCS); Fixed Femur - Moving Tibia (FF-MT). To the tibia moving on the flexing femur; (b) Joint flexion-extension rotation is applied to the femur and the remaining five dependent movements are applied to the tibia part; Hinged Femur - Perpendicular Moving Tibia (HF-PMT).

Hence, this complex collective set of motions is called 'Hinged Femur - Perpendicular Moving Tibia' (HF-PMT) for convenience. As a result, tibia motion is defined with respect to the femur. The origins of the femoral and GCS frames and their  $X$ -axes are common. Consequently, angular displacement in the sagittal plane is femoral flexion-extension rotation about the  $X_F$ -axis, while the other two dependent rotations and three dependent translations are observed on the tibia with respect to the GCS ( $X_{Ti}$ ,  $Y_{Ti}$ ,  $Z_{Ti} \Rightarrow X_{Tf}$ ,  $Y_{Tf}$ ,  $Z_{Tf}$ ). Passive flexion movement is achieved by only exerting flexion-extension rotation on the femur (i.e.  $X_F$ ,  $Y_{Fi}$ ,  $Z_{Fi}$  rotated by alpha ( $\alpha$ ) degrees to  $X_F$ ,  $Y_{Ff}$ ,  $Z_{Ff}$ ). The remaining five degrees of freedom occur on the tibia with respect to the GCS. The original motion defined with 6 DoF of the joint in Figure 2.18a remains the same except the reference of rotation is redefined for perception and compliance with Grood and Suntay. The motions in Figure 2.18a and Figure 2.18b are identical.

### 2.5.5 Knee laxity

Clinically, radiological modalities such as Computed Tomography (CT) and Magnetic Resonance Imaging (MRI) are the most commonly used methods to diagnose soft tissue injuries, including ligament tears. However, these imaging techniques are not sufficient to accurately assess the extent or severity of the tear [77, 78, 79]. When the mechanical condition of the injured knee permits (e.g., minimal swelling) and the patient's pain is manageable, a manual clinical evaluation is essential for diagnosing anterior cruciate ligament (ACL) or other ligament injuries [80]. Physical examinations – such as ligament laxity assessments, including the drawer test – are routinely performed to determine whether ligaments are torn or ruptured. Moreover, these tests provide clinically valuable benchmarks for comparing knee joint laxity before and after surgical ligament reconstruction [81, 82].

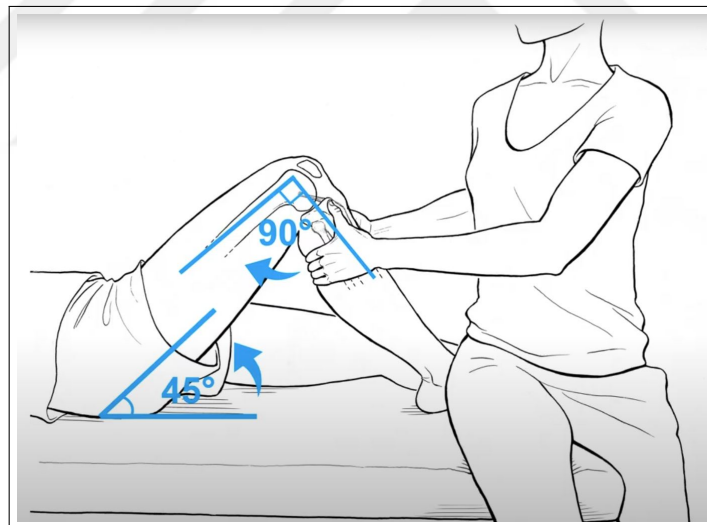


**Figure 2.19** The anterior drawer and posterior drawer tests are to check integrity of the Anterior Cruciate Ligament (ACL) and Posterior Cruciate Ligament (PCL) in the knee. The patient lies supine with the hip flexed 45 degrees and the knee flexed 90 degrees and hamstrings are relaxed. Examiner grasps the proximal tibia with both hands just below the joint line of the knee. The excessive translation of tibia anteriorly and posteriorly indicates ACL and PCL rupture respectively. Illustrations were adapted from openly accessible content published by Yost [19].

The anterior drawer test is commonly used to assess the integrity of the anterior cruciate ligament (ACL) in the knee. It is considered both a reliable and practical procedure for diagnosing ACL tears. Clinically, the patient lies supine on an examination table with the tested leg positioned at 90 degrees of knee flexion and 45 degrees of hip

flexion (Figure 2.19). The clinician typically sits on the patient's forefoot to stabilize the foot and prevent it from sliding. At 90 degrees of knee flexion, the ACL aligns nearly parallel to the anterior axis of the tibia. It is essential that the clinician verifies if the hamstring muscles is relaxed before proceeding. Provided all conditions, the tibia is drawn anteriorly. The examiner grasps the proximal tibia with both hands just below the joint line. If the contralateral (unaffected) knee is healthy, it is often used for comparison. In the absence of joint hyperlaxity, anterior tibial translation is normally less than 5 mm. A positive test is indicated by increased anterior tibial displacement: grade I for approximately 5 mm, grade II for 5 - 10 mm, and grade III for displacement greater than 10 mm. This method is also implemented in the computational model (CM) to simulate ACL loading and evaluate both pre- and post-operative knee joint behavior.

The posterior drawer test is used to assess the integrity of the posterior cruciate ligament (PCL) in the knee. Similar to the anterior drawer test, the patient lies supine with the hip flexed to 45 degrees and the knee flexed to 90 degrees, ensuring the hamstring muscles are relaxed, as illustrated in Figure 2.19. The examiner sits on the forefoot of the involved leg to stabilize it and grasps the proximal tibia with both hands just below the knee joint line. A firm posterior force is then applied to the anterior surface of the tibia. A positive test is indicated by the absence of a firm end-feel or by excessive posterior translation of the tibia compared to the uninvolved side, which may suggest a rupture of the PCL.

The valgus stress test is used to assess the integrity of the medial collateral ligament (MCL) of the knee. The patient lies supine with the legs relaxed, the hip abducted, and the knee flexed between 20 and 30 degrees, as shown in Figure 2.20. The examiner places one hand over the lateral joint line of the knee and the other on the medial aspect of the ankle. A valgus force – directed away from the body's midline – is applied through the ankle, in alignment with the plane of the ligament. While a small degree of medial joint gapping is considered normal, pain or excessive gapping is indicative of a positive test, suggesting injury to the MCL.

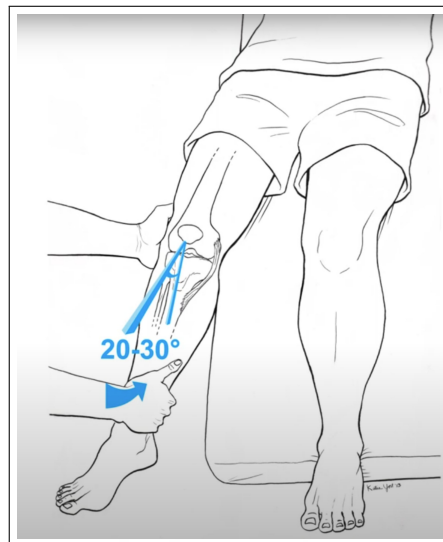


**Figure 2.20** The valgus stress test is to test integrity of the Medial Collateral Ligament (MCL) in the knee. The patient lies supine with the legs relaxed, the hip abducted, and the knee flexed 20 to 30 degrees. A valgus force pushing away from midline is applied through the ankle in the same plane as the ligament. Some gapping is normal but pain or excessive gapping is a sign of a positive test. This indicates injury to the MCL. Illustrations were adapted from openly accessible content published by Yost [19].

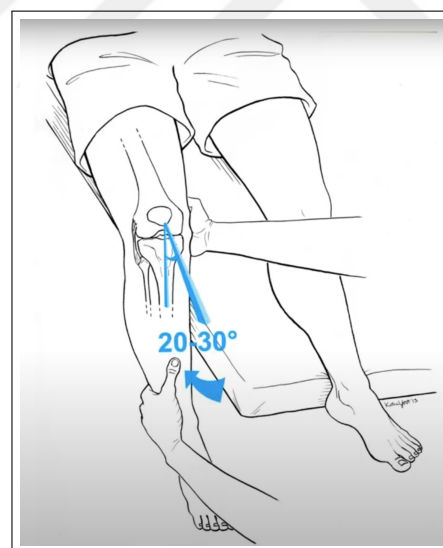


**Figure 2.21** The varus test is used to assess damage to the LCL and the ALL in the knee. The patient lies supine with the legs relaxed, the hip abducted, and the knee flexed 20 to 30 degrees. A varus force pushing towards midline is applied through the ankle in the same plane as the ligament. Some gapping is normal but pain or excessive gapping is a sign of a positive test. This indicates injury to the LCL and the ALL. Illustrations were adapted from openly accessible content published by Yost [19].

The varus test is used to assess damage to the LCL and the ALL in the knee. The patient lies supine with the the legs relaxed, the hip abducted, and the knee flexed

20 to 30 degrees (Figure 2.21). The examiner places one hand over the medial joint line of the knee and the other hand on the lateral aspect of the ankle. A varus force pushing towards midline is applied through the ankle in the same plane as the ligament. Some gapping is normal but pain or excessive gapping is a sign of a positive test. This indicates injury to the LCL and the ALL.

The varus stress test evaluates the lateral collateral ligament (LCL) and the anterolateral ligament (ALL). The patient is positioned similarly: supine with the legs relaxed, the hip abducted, and the knee flexed between 20 degrees and 30 degrees, as shown in Figure 2.21. The examiner places one hand on the medial joint line of the knee and the other on the lateral aspect of the ankle. A varus force – directed toward the body’s midline – is applied through the ankle, in the same plane as the ligament. As with the valgus test, minor lateral gapping can be normal; however, the presence of pain or excessive gapping indicates a positive test and potential injury to the LCL and/or ALL.

### 2.5.6 Parametric modelling

In the experimental study, marker positions of NK were sampled at 100 Hz as passive knee motion was exerted manually [5, 6]. During passive knee flexion, the remaining five motions - internal-external, varum-valgum, medial-lateral, anterior-posterior, and superior-inferior - are coupled with the flexion movement. Therefore, all these dependent movements can be parameterized with respect to flexion-extension rotation. In NK, the knee was flexed to a maximum flexion starting from the full-extension position and then extended back to near the starting position, forming a full passive knee movement cycle. The repository contains four such cycles consecutively. The same motion parameterization is established in CM for the same four cycles. This establishes NK as the gold standard for validating our proposed model (CM).

The passive movement of the knee was repeated four times during the NK setup. Since passive knee movement depends on flexion-extension (FE) and exhibits a periodic

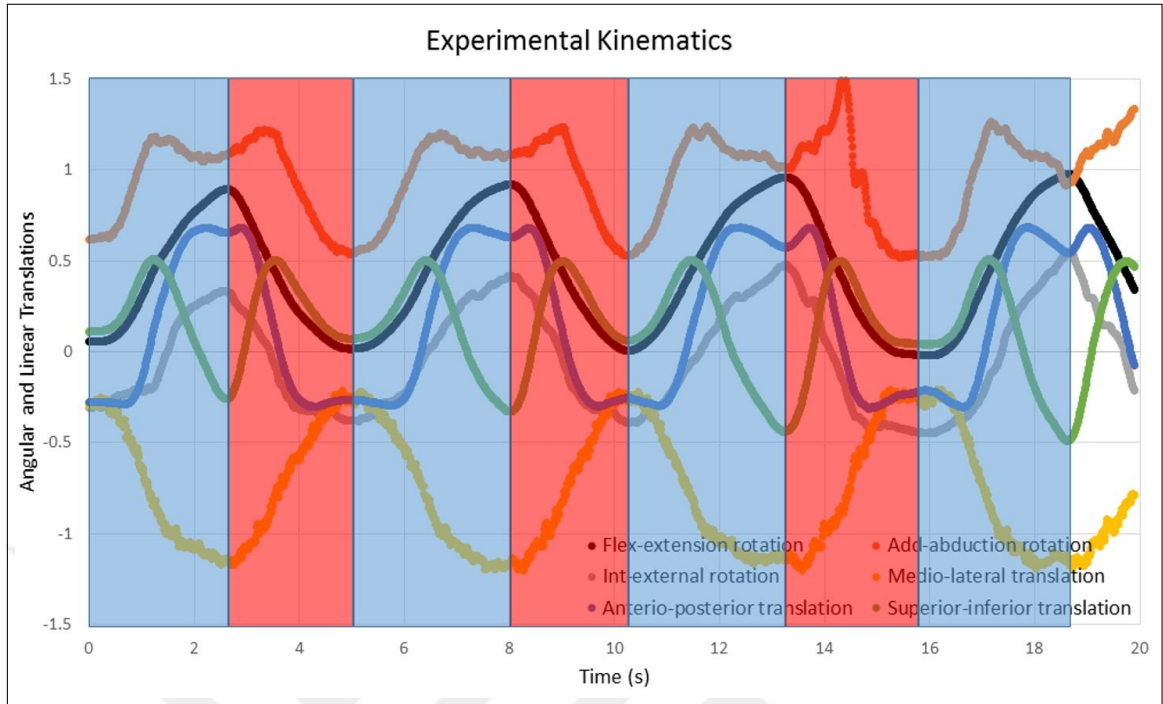


**Figure 2.22** Graphs represent the means and variations of four consecutive trajectories recorded in NK. During four consecutive passive knee flexions, five dependent displacements - abduction-adduction (red), internal-external (purple), medial-lateral (yellow), anterior-posterior (blue), and superior-inferior (green) - are recorded and graphed in colors. The abscissa is an independent flexion-extension motions in NK passive knee flexion. The mean curves of five dependent motion (passive-flexion components) for experimental kinematics are transformed from joint coordinate axis to the Hinged Femur - Perpendicular Moving Tibia (HF-PMT) system. Red (abduction-adduction) and purple (internal-external) colored lines represent rotation components in degrees, while lines represent translation components in millimeters.

nature with respect to FE, we chose to parametrize the data accordingly. All kinematic data - flexion-extension (black), abduction-adduction (red), internal-external (purple), medial-lateral (yellow), anterior-posterior (blue), and superior-inferior (green) - are represented graphically in Figure 2.22. The figure also shows the division of data into periods, with blue and red shaded areas indicating the flexion and extension phases of passive knee movement, respectively. Note that the final extension phase is not included, as it does not constitute a full period, corresponding to the full extension of the knee joint. As a result, four flexion and three extension movements were segmented based on the maximum and minimum values of FE (Figure 2.22). The peak flexion (minimum FE value) and peak extension (maximum FE value) were not consistent across cycles, with peak flexion values recorded as  $112.85^\circ$ ,  $116.28^\circ$ ,  $121.35^\circ$ , and  $123^\circ$  and peak extension values as  $6.81^\circ$ ,  $1.72^\circ$ ,  $0.37^\circ$ , and  $-2.99^\circ$ . Data beyond  $112^\circ$  of

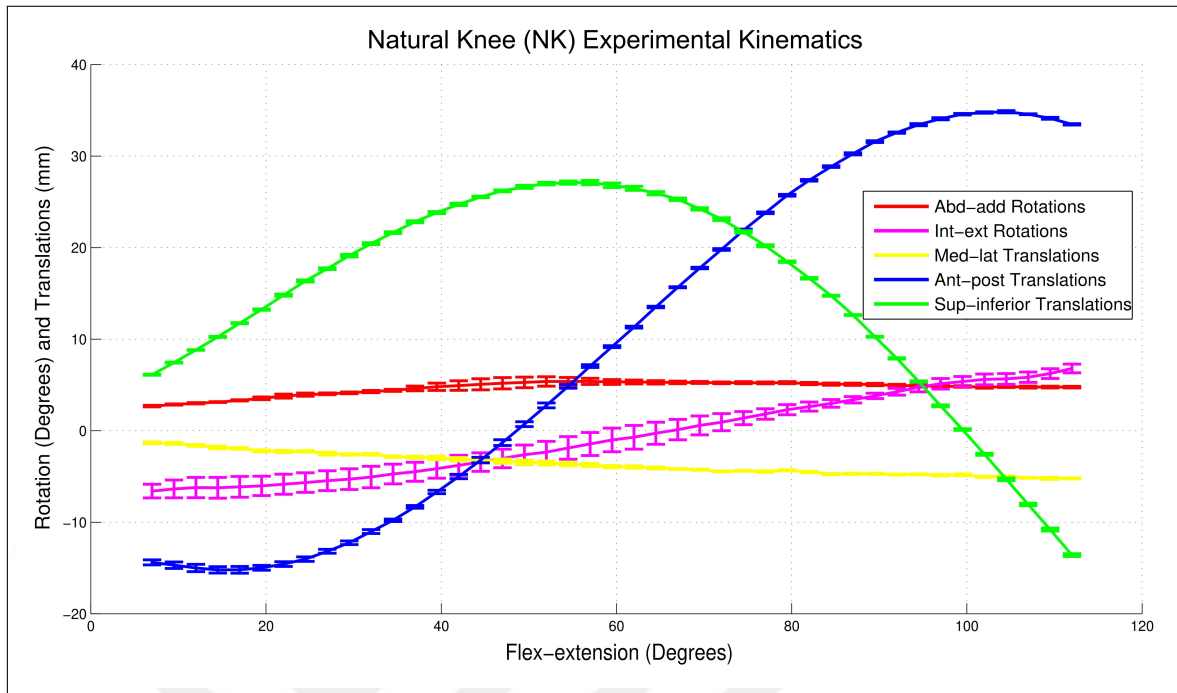


**Figure 2.23** Graphs represent the means and variations of four consecutive trajectories recorded in NK. During four consecutive passive knee flexions, five dependent displacements - abduction-adduction (red), internal-external (purple), medial-lateral (yellow), anterior-posterior (blue), and superior-inferior (green) - are recorded and graphed in colors. The abscissa is an independent flexion-extension motions in NK passive knee flexion. The mean curves of five dependent motion (passive-flexion components) for experimental kinematics are transformed from joint coordinate axis to the Hinged Femur - Perpendicular Moving Tibia (HF-PMT) system. Red (abduction-adduction) and purple (internal-external) colored lines represent rotation components in degrees, while lines represent translation components in millimeters.

flexion and 7° of extension were truncated. Finally, the parametrization of the five dependent movements with respect to FE was resampled.

Figure 2.23 illustrates the mean of the four flexion and three extension passive knee movements parametrized by FE. The five dependent displacements - abduction-adduction (red), internal-external (purple), medial-lateral (yellow), anterior-posterior (blue), and superior-inferior (green) - are displayed in color for clarity.

### 2.5.7 Quantitative comparison; NK vs. CM

For each of the five movements described in the Section above (2.5.5), CM simulated (HF - PMT) and NK experimental (HF - PMT) kinematics are pairwise compared using linear regression analysis, based on:

$$y = b_0 + b_1x \quad (2.6)$$

where  $y$  represents the experimental data and  $x$  represents the simulated data. The model calculates linear relationship parameters:  $b_0$  (*intercept*),  $b_1$  (*slope*), Root Mean Squared Error (*RMSE*), fitting quality (adjusted *R-squared*), and statistical significance for estimation (*p-value*). If the experimental outputs are similar to the simulated kinematics the regression parameters are expected to be close to one for *slope* and *R-squared*, close to zero for *RMSE* and *intercept*, and less than 0.05 for *p-value*.

### 3. RESULTS

#### 3.1 Kinematic results

The earlier model with 14 bundle without ALL have instability at high flexion. Specifically, internal rotation abruptly escalates resulting in abnormal slope of the internal-external kinematics as shown orange graph in Figure 3.1. Referring to Chapter 2, Section 2.4.10, Antero-Lateral Ligament (ALL) is suspected to cause that issue at early stages where it does not included in the Computational Model (CM).

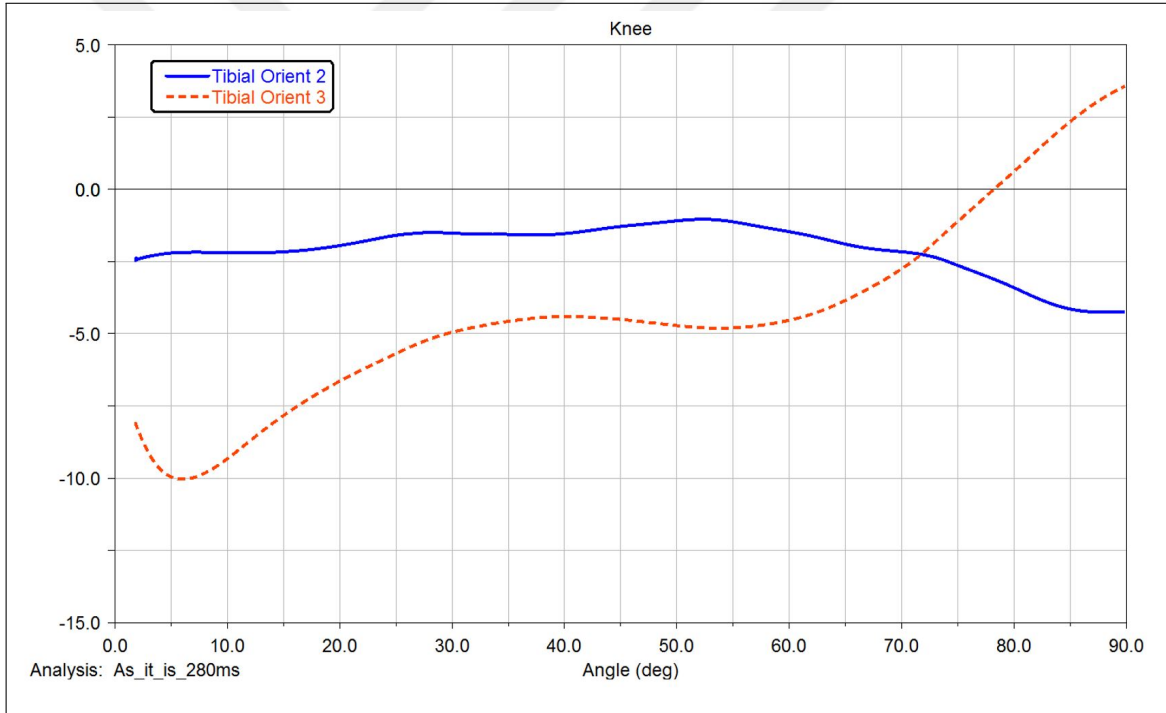


**Figure 3.1** Mean curves and standard deviations of passive-flexion components for simulated kinematics (CM) transformed from joint coordinate axis to the Hinged Femur - Perpendicular Moving Tibia (HF-PMT) system. Rotation components in degrees are represented by red (abduction-adduction) and purple (internal-external) colored lines. Translation components in millimeters are depicted by yellow (medial-lateral), blue (anterior-posterior), and green (superior-inferior) lines.

Recent research studies conveyed on cadaver show that at initial flexion angles (<30 degrees) the internal rotation strain is loaded mainly at cruciate ligaments, beyond 30 degrees anterolateral ligament has observed more involved to internal rotation

stability [83]. In the model, we also have joint subluxation during higher flexion angles of the CM. Therefore we include in the CM model.

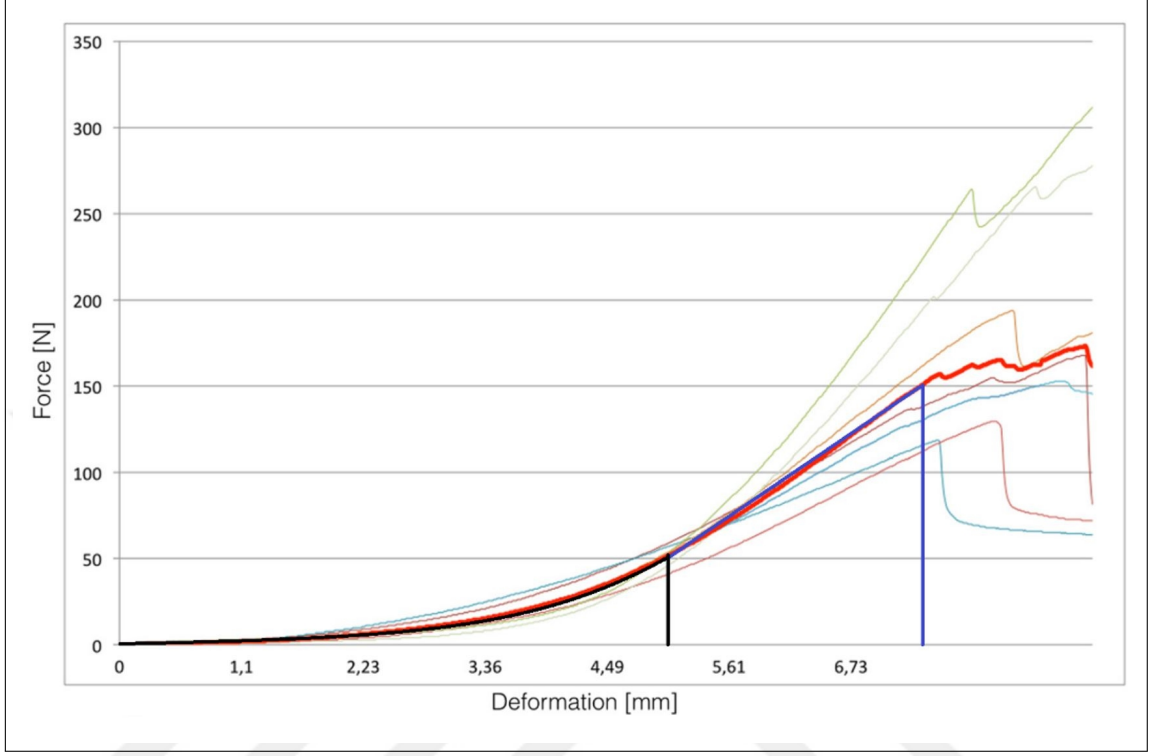


**Figure 3.2** Anterolateral ligament (ALL) force-deformation curves [20]. Red curve is mean of all experimental samples. It divided into blue thick line which is linear region and black thick line which is quadratic region.

Several generic force-elongation equation of the ligament available in the literature, and they are consistent. However, few studies are performed on mechanical properties of the ALL ligament since anatomy studies proved its existence and the prevalence of the ligament is high. Mechanical properties study by Helito et. al. taken as reference for obtaining quadratic stiffness and linear region stiffness for anterolateral ligament [20]. To engage quadratic coefficient ( $K_Q$ ), Matlab is used for fitting black curve (quadratic region) as illustrated on the Force-Deformation curve (Figure 3.2).

$$F_{ALL} = \begin{cases} 0 & \Delta L_{ALL} < 0 \\ K_{QALL}(\Delta L_{ALL})^2 & 0 \leq \Delta L_{ALL} \leq 5.05 \\ K_{LALL}(\Delta L_{ALL} - \epsilon_{1ALL}L_{0ALL}) & 5.05 < \Delta L_{ALL} \end{cases} \quad (3.1)$$

Force transition from quadratic to linear region observed when deformation is 5.05.

Stiffness coefficient for the quadratic region is  $K_{QALL}=1.9606$ , and linear stiffness is  $K_{LALL}=33.8983$ . The slack length of ALL is  $L_{0ALL}=47.685$  and  $\epsilon_{1ALL}$  is calculated as 0.075 for smooth transition to linear region using MATLAB code in Appendix B.

The resulting five dependent motions, as described in Chapter 2, Section 5.4, due to Passive Knee Flexion (PKF) are shown in Figure 2.23 and Figure 3.3 for NK and CM, respectively.

For each of the motions, there are four consecutive recordings in NK. The trajectories shown in Figure 2.23 belong to the averages and variations of the four trajectories.

The flexion-extension data in NK is also used to drive knee joint simulations in CM. Consequently, there are also four flexion and extension trajectories in CM, resulting in four trajectories for the five dependent joint motions. The average and variance of the resulting trajectories in CM are shown in Figure 3.3.

In Figure 3.4 and Figure 3.5, six trajectories corresponding to passive knee flexion are pairwise compared for experimental (NK) and simulated (CM) motions.

Finally, Table 3.1 comprises the results of two ranges of motion; a) truncated (7-90 degrees) and b) higher flexion range (7-112 degrees) flexion-extension movements. The regression analysis indicate experimental (NK) and simulated (CM) trajectories are highly correlated and the results are statistically significant, with a p-value less than 0.05 for all paired kinematics.

Figure 3.4 comprises six graphs, each of which corresponds to one motion component of knee passive flexion. The two curves in each figure represent the average of four trajectories for NK (red) and CM (blue), with the corresponding variances.

Figure 3.4a shows flexion-extension rotation with respect to itself. It is illustrated for completeness of the argument for all the six knee joint motions. This independent

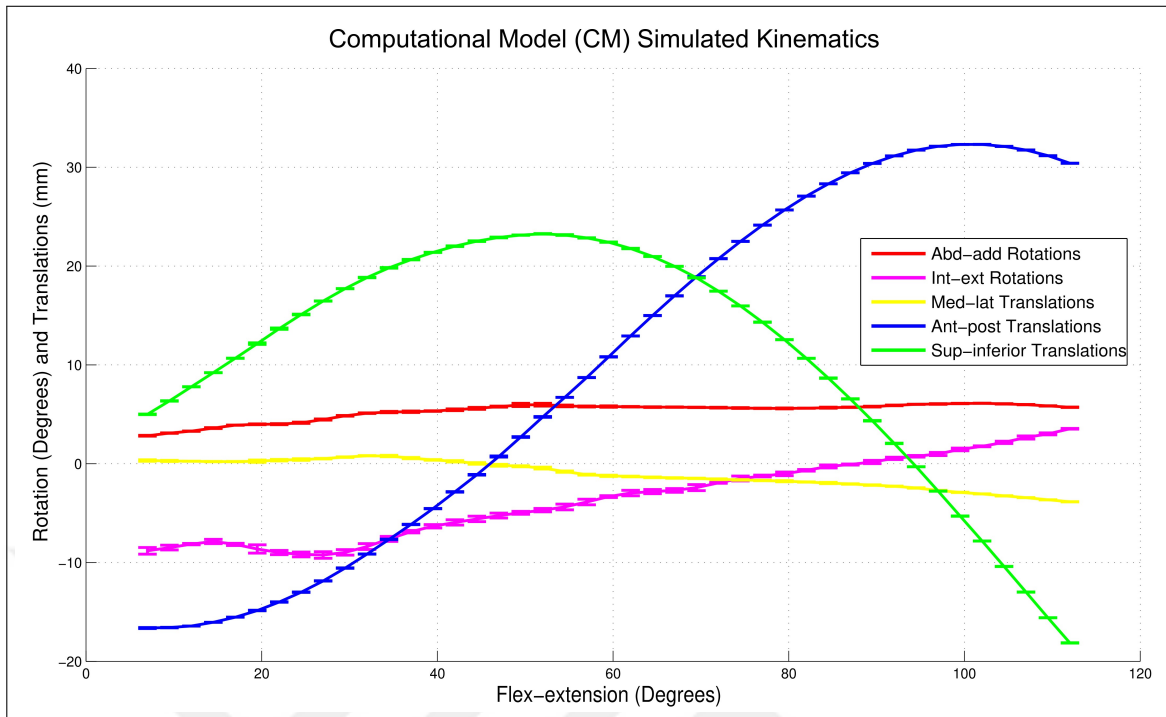


**Figure 3.3** Mean curves and standard deviations of passive-flexion components for simulated kinematics (CM) transformed from joint coordinate axis to the Hinged Femur - Perpendicular Moving Tibia (HF-PMT) system. Rotation components in degrees are represented by red (abduction-adduction) and purple (internal-external) colored lines. Translation components in millimeters are depicted by yellow (medial-lateral), blue (anterior-posterior), and green (superior-inferior) lines.

motion is linearly sampled and drives the rest of the five dependent joint motions for both NK and CM experiments.

Figure 3.4b depicts the trajectory for abduction-adduction of the knee joint with respect to the flexion-extension rotation for both NK(red) and CM(blue) for the four repetition. The abduction-adduction trajectory resulting from four executions shows noticeable variation between 40 to 60 degrees of flexion-extension. Although it is not as high as in NK, abduction-adduction in CM also varies more noticeably between 40 to 60 degrees of flexion-extension. While average trajectories are not exactly superimposed for NK and CM, the two trajectories are highly correlated as can be seen in Figure 3.5b and Table 3.1.

The third angular displacement, internal-external rotation, has a close-to-linear relation with flexion-extension. While CM internal-external rotation varies less across

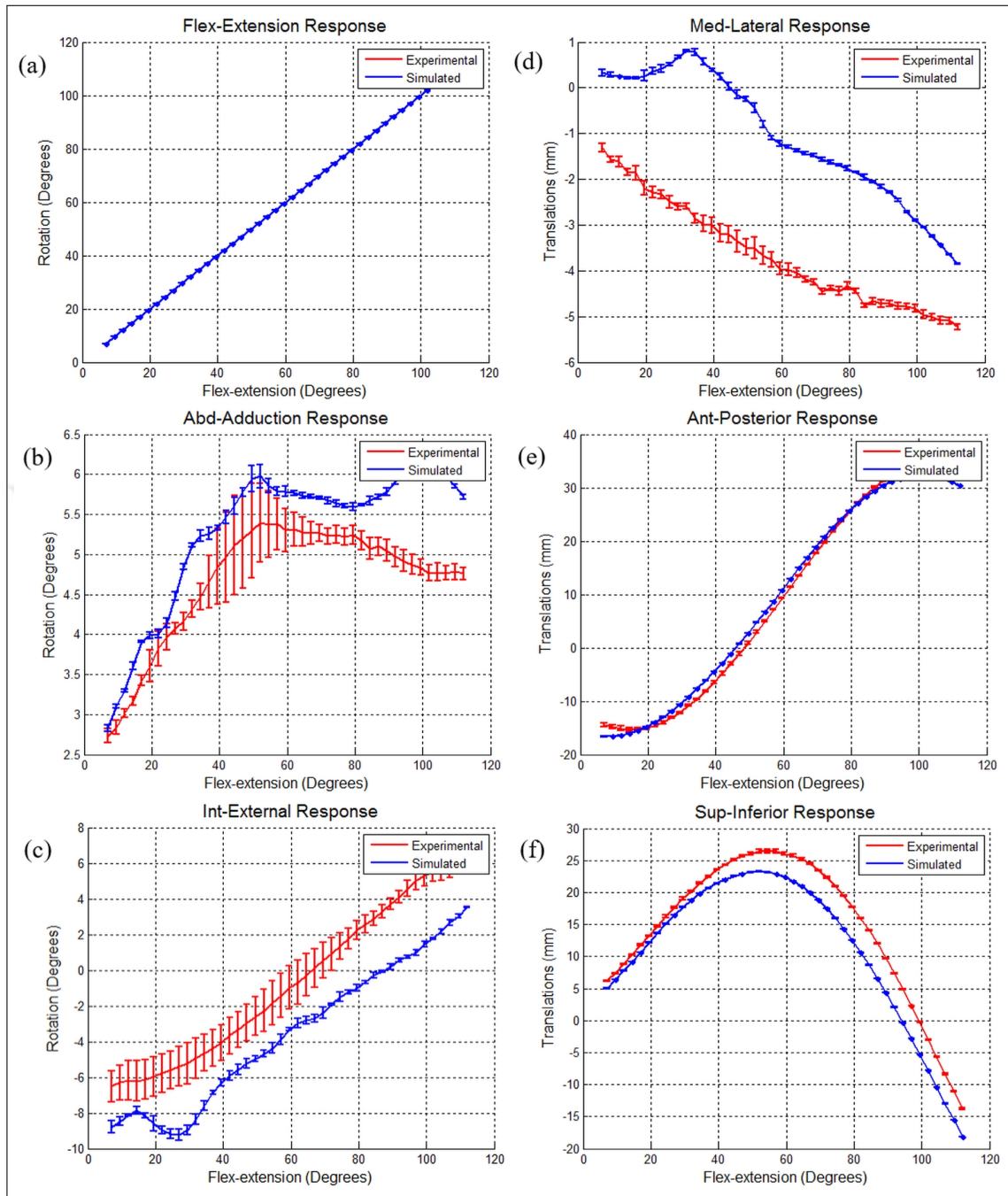


**Figure 3.4** Experimental and simulated results in Figure 2.23 and 3.3 are reillustrated pairwise for the comparison of NK (red) and CM (blue) kinematics. In the first row, angular displacements are compared: (a) flexion-extension (input motion), (b) abduction-adduction, and (c) internal-external rotations. In the second row, linear displacements are compared: (d) medial-lateral, (e) anterior-posterior, and (f) superior-inferior translations, all in response to NK flexion-extension. Figure (a) shows the comparison of flexion-extension by itself, where a straight line is expected. It is presented here for the completeness of the argument.

experiments, as shown in Figure 3.4c (blue), the NK internal-external curve (Figure 3.4c red) is smoother. NK and CM internal-external rotations are highly correlated,

as shown in Figure 3.5c.

The medial-lateral displacement is inversely proportional to flexion-extension. The medial-lateral displacement has the least excursion of all translational displacements. The experimental and simulated medial-lateral displacements do not superimpose, however their offset is less than two millimeters throughout the experiments. CM medial-lateral displacement (Figure 3.4d, blue) shows less variance accross the entire motion compared to NK (Figure 3.4d, red) with a wider variation in the range of 40 to 60 degrees of flexion-extension. Correlation between NK and CM, although not as high as other dependent motions, is still significant, as shown in Figure 3.5d and Table 3.1.

Forward and backward movement of the tibia along the floating axis F (anterior-posterior displacement) is almost identical for NK (red) and CM (blue), as shown in Figure 3.4e. The similarity throughout the entire range of flexion-extension between NK and CM is highest for anterior-posterior displacement, as illustrated in Figure 3.5e, producing the highest correlation in Table 3.1.

Figure 3.4f illustrates mean trajectories of superior-inferior displacements for measured NK (red) and compiled CM (blue) experiments during passive knee flexion. Although there are some discrepancies at higher flexions, the resulting curvatures are closely related, producing significant correlation results as shown in Figure 3.5f and Table 3.1.

Regression results for frontal plane rotation (abduction-adduction) are better in the 7-90 degrees range than in the 7-112 degrees ranges, except for the *intercept*.

The external-internal rotation similarity comparison parameters, *RMSE*, and *R-squared* values are slightly better for 7-112 degrees, whereas *slope* and *intercept* values are closer to ideal for 7-90 degrees of flexion-extension.

For mediolateral translation, *slope* and *R-squared* parameters for 7-112 degrees,

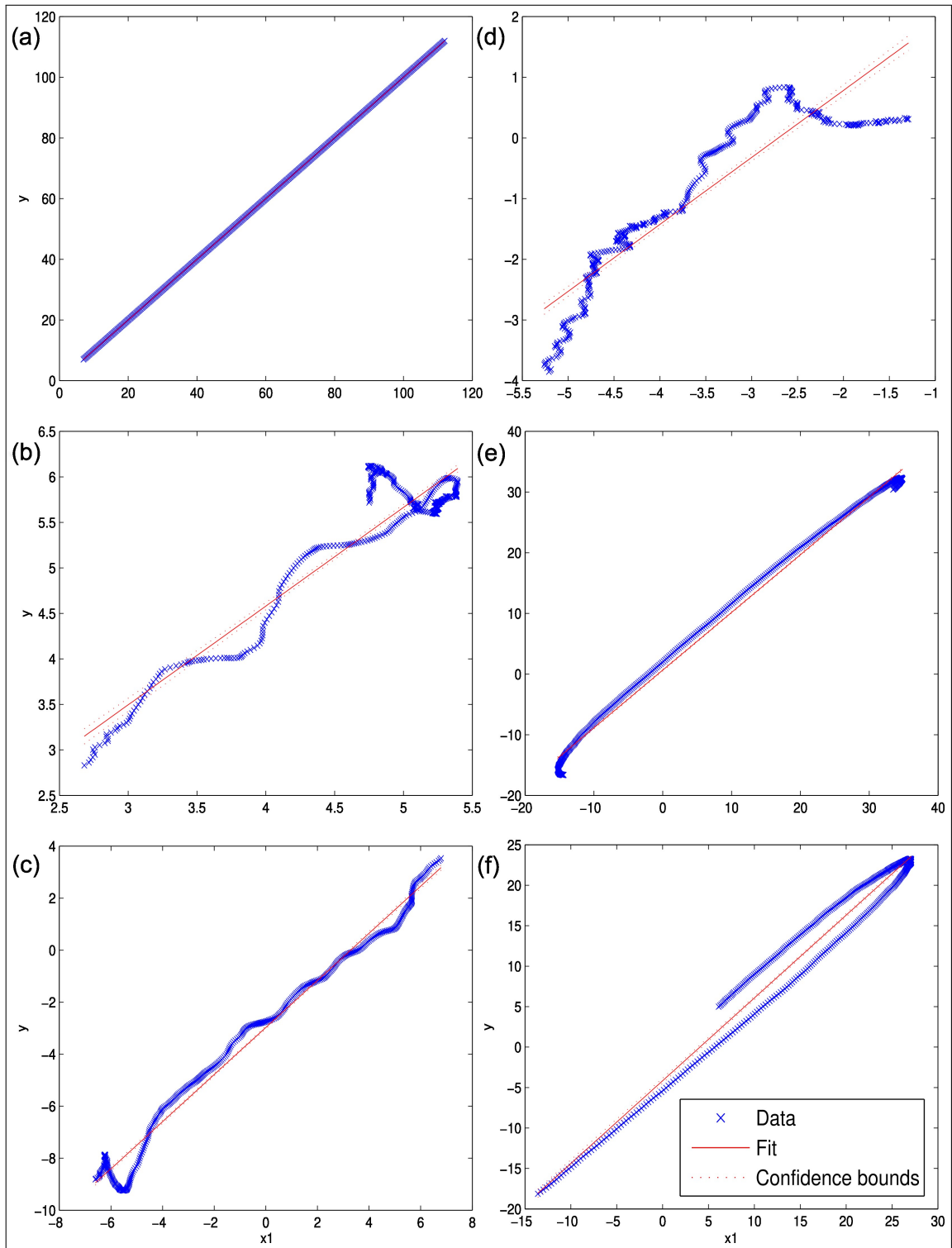


**Figure 3.5** Linear regression model plots for comparing the similarity of experimental NK and simulated CM outcomes of passive knee flexion. The left column contains angular kinematics: (a) flexion-extension, (b) abduction-adduction, (c) external-internal rotations. The right column consists of linear kinematics: (d) medial-lateral, (e) anterior-posterior, and (f) superior-inferior translations.

**Table 3.1**  
Linear Regression Analysis on Similarity of NK and CM.

|           | Slope |       | Intercept |        | RMSE  |       | R-squared |       |
|-----------|-------|-------|-----------|--------|-------|-------|-----------|-------|
|           | 90°   | 112°  | 90°       | 112°   | 90°   | 112°  | 90°       | 112°  |
| <b>FE</b> | 1.000 | 1.000 | 0.000     | 0.000  | 0.000 | 0.000 | 1.000     | 1.000 |
| <b>AA</b> | 1.048 | 1.084 | 0.276     | 0.243  | 0.148 | 0.302 | 0.971     | 0.875 |
| <b>EI</b> | 0.934 | 0.905 | -2.862    | -2.973 | 0.522 | 0.510 | 0.971     | 0.983 |
| <b>ML</b> | 0.866 | 1.104 | 2.340     | 2.990  | 0.439 | 0.540 | 0.797     | 0.839 |
| <b>AP</b> | 0.992 | 0.950 | 0.862     | 0.653  | 1.230 | 1.420 | 0.994     | 0.994 |
| <b>SI</b> | 0.889 | 1.023 | -1.215    | -4.157 | 1.780 | 1.850 | 0.907     | 0.973 |

and *intercept* and *RMSE* parameters for 7-90 degrees are marginally better.

The anterior-posterior translation similarity calculations resulted in an *R-squared* value of 0.994 for both 7-90 and 7-112 degrees of flexion-extension. For 7-90 degrees, the *slope* and *RMSE* parameters are closer to ideal, one. The *intercept* is better for 7-112 degrees of flexion-extension.

When comparing NK superior-inferior translations with the results of CM, the *slope* and *R-squared* values are better for 7-112 degrees, while *intercept* and *RMSE* values are better for 7-90 degrees of flexion-extension.

### 3.2 Knee laxity results

In this study, the anterior drawer test is simulated to replicate the clinical procedure typically performed by an examiner. In clinical practice, the examiner sits on the patient's forefoot to stabilize the foot while allowing the ankle joint to remain unconstrained. This setup is modeled in the computational model (CM) by defining a hinge joint between the tibia and the ground at the center of the ankle joint, denoted as *Cank*. This hinge allows for dorsiflexion and plantarflexion of the ankle.

During the test, the patient's knee is positioned at 90 degrees of flexion, as illustrated in Figure 2.19. To achieve this configuration in the CM, the femur is rotated 90 degrees clockwise, resulting in a hip joint angle of 45 degrees flexion as shown in Figure 2.19. This configuration is verified by measuring the angle between the femur and the ground, confirming 45 degrees of hip flexion.

The anterior force applied by the examiner during the drawer test is simulated by applying a point force to the origin of the tibia,  $O_{tib}$ , in the anterior direction along the tibial Y-axis,  $\mathbf{Y}_T$ .

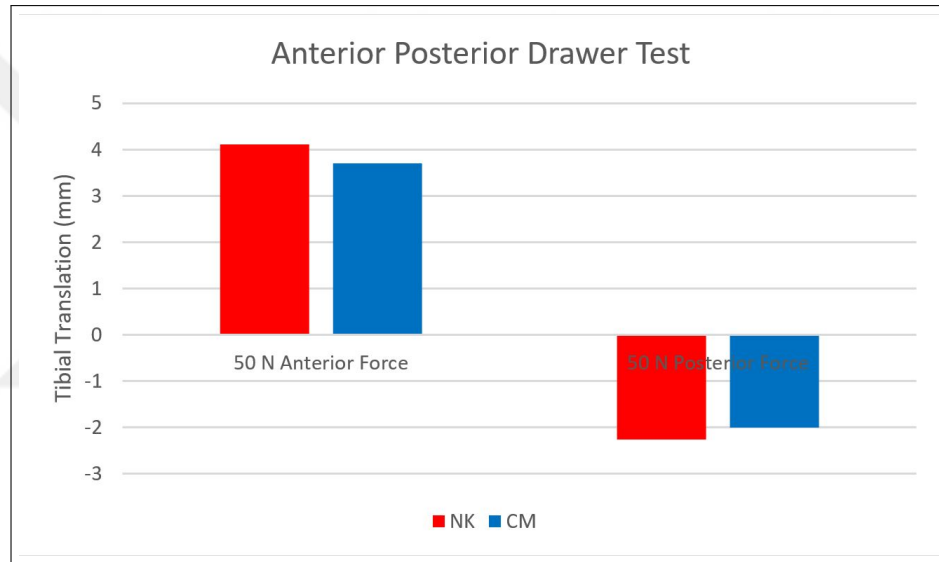


**Figure 3.6** Anterior and posterior drawer test for Natural Knee (NK) and Computational Model (CM).

The anterior drawer test is implemented in the computational model (CM) by applying a 50 N anterior force at a loading rate of 10 N/s. Under this applied force, the model reached equilibrium after translating 3.707 mm in the anterior direction (Figure 3.6). In comparison, the anterior translation measured in the Natural Knee (NK) experiment was 4.113 mm. When the direction of the 50 N force was reversed (i.e., applied posteriorly), the tibial translation in the CM was -1.995 mm, whereas the corresponding posterior translation in the experimental knee is -2.260 mm, as shown in Figure 3.6.

The valgus and varus stress tests were also implemented in the computational

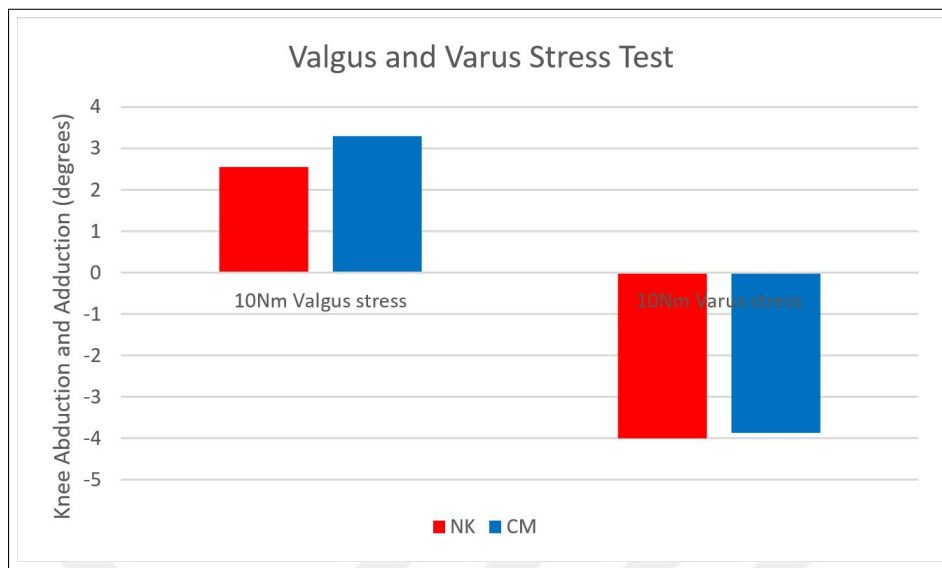


**Figure 3.7** Valgus and varus stress results for Natural Knee (NK) and Computational Model (CM).

model (CM). The knee joint was positioned at 30 degrees of flexion, replicating the clinical examination setup (Figures 2.20 and 2.21). When a valgus torque of 10 Nm was applied at a rate of 1 Nm/s, the CM produced 3.295 degrees of abduction. In comparison, the abduction measured in the Natural Knee (NK) experiment was 2.545 degrees, as shown in Figure 3.7.

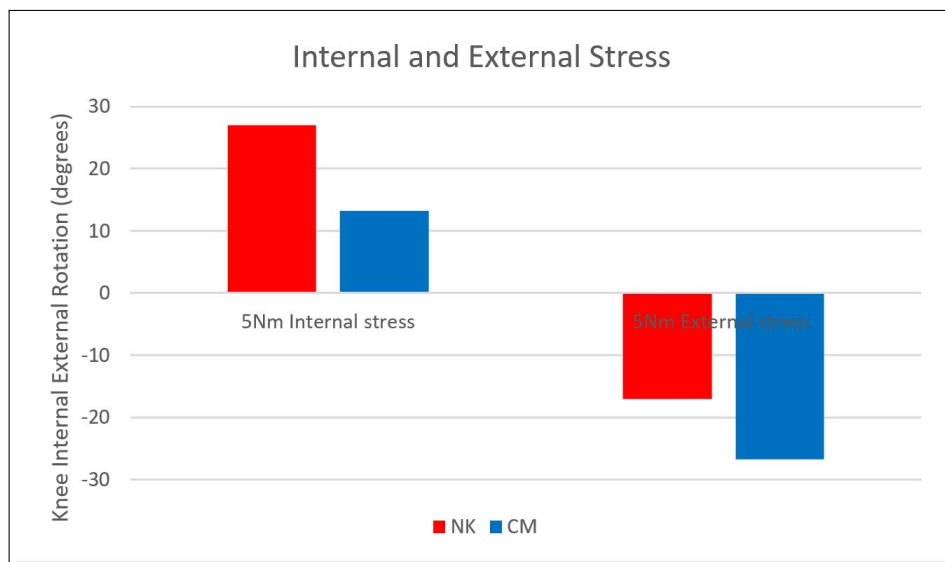


**Figure 3.8** Internal and external stress results for Natural Knee (NK) and Computational Model (CM).

Clinically, internal and external rotational stress tests are rarely performed.

However, for the comprehensiveness of the laxity evaluation, it was assumed that the patient lies in a prone position with the involved knee flexed to 90 degrees – similar to the positioning used in thigh-foot angle assessment. Subsequently, internal and external torques of 5Nm (applied at a rate of 1Nm/s) were applied to the tibia. The computational model (CM) responded with 13.277 degrees of internal rotation and 26.729 degrees of external rotation of the knee joint (Figure 3.7). In comparison, the Natural Knee (NK) measurements showed 26.988 degrees of internal and 16.981 degrees of external rotation.

Finally, to simulate the pre-operative condition before anterior cruciate ligament (ACL) reconstruction, the CM is modeled with a complete ACL rupture by disabling both aACL and pACL elements. This resulted in 5.786mm of anterior tibial translation.

## 4. DISCUSSION and CONCLUSION

In conclusion, the simulated model produces a similar response to the actual knee when compared under the generally accepted conditions of passive knee flexion.

In the proposed computational model a multi-body tibiofemoral joint is constructed including *subject-specific* anatomical structures of the joint. Solid shell structures are extracted from patient MR images for solid parts (bone) and for contact surfaces of cartilages. Ligament insertion sites are also defined from MR images based on anatomical relevances. A visual comparison in Figure 2.23 and Figure 3.3 shows that the joint trajectories produced in NK and CM are similar for the same flexion-extension for the input motion. The computational model response is within an acceptable range when individual joint parameters are compared pairwise for NK and CM, as illustrated in the Figure 3.4. This comparison approach aligns with the method used in previous studies in the literature to demonstrate model validity [48, 50, 49]. Additionally, we presented a quantitative method to compare actual and computational model responses where the results are shown in Table 3.1. Pearson's correlation results between the responses show a strong correlation ( $r > 0.89$ ) for each of the five dependent motions. In this study, a linear regression model of experimental kinematics is fitted to the simulated kinematics (Figure 3.5). In summary, regression results, RMSE and R-squared show that the experimental (NK) and simulated (CM) motions are acceptably similar. Quantitative similarity parameters allow us to use our model with confidence in patient-specific studies. This approach allows us to customize the proposed model based on the parameters acquired from the knee joint MR images of the patient.

Knee joint laxity is commonly assessed by applying external forces to the tibia in directions other than flexion. Theoretically, during passive knee flexion, at each increment of the flexion angle, forces (anterior-posterior, medial-lateral, and superior-inferior) and moments (valgus-varus and internal-external) should be applied and recorded. However, performing such a comprehensive laxity assessment is computation-

ally impractical and unrealistic. Moreover, it does not necessarily provide meaningful additional insight into the functional state of the knee joint.

Previous studies have used six-degree-of-freedom robotic systems and force sensors to apply external tibial loads at discrete knee flexion angles (0, 15, 30, 60, 90 and 120 flexion degrees) to investigate joint laxity [84, 85]. In studies exploring knee injury mechanisms, researchers have focused on ligament-specific laxity assessments [86, 87].

In the present study, we primarily simulated clinically performed joint laxity tests. In addition to conventional clinical assessments, we also investigated the response of the knee to internal and external rotational stresses. The outcomes of our computational model align well with both experimental data from the Natural Knee (NK) dataset and previously published literature [48, 50, 49, 86, 84, 85]

In this study a 3D model platform of the tibiofemoral joint is established. Although patella and meniscus do not contribute significantly to the passive knee motion they will become important when other modes of knee joint motion are investigated. Therefore patella and meniscus will be included in the future for a more comprehensive model. Furthermore, structural and functional properties will be refined to better represent realistic soft tissue behaviors, such as cartilaginous contact and ligamentous forces.

The proposed model's capacity extends to simulating knee pathologies, such as ligament rupture and cartilage degeneration, allowing for the evaluation of surgical modifications in a quantitative manner, assessing regression or progression. Surgeons accustomed to traditional methods, i.e. the Drawer test, for ligament assessment and treatment evaluation, could benefit from performing such tests in simulation environment using the presented tibiofemoral model.

We demonstrated that using MR images of lower leg and knee is sufficient for the reconstruction of an acceptable computational tibiofemoral model customized to a patient. With this approach, although it may increase preparation time, surgeons will

have the opportunity to study their approach prior to operation in detail, accounting for several possible outcomes.

Our future studies involve developing an automated predictive tool for post-ligament reconstruction surgery based on presented model.

Ultimate aim of the study is to model the whole human gait cycle implementing ankle and hip joints together with the knee model. This will allow us to study the impact of surgical procedures on walking and balance prior to surgeries.



## APPENDIX A. OVERVIEW OF REVIEWED KNEE MODEL STUDIES

**Table A.1**  
Summary of studies on a knee model.

| Study           | Dim | RoM(°) | Bone G | Mod    | Lig# | Lig | Car | Men |
|-----------------|-----|--------|--------|--------|------|-----|-----|-----|
| Wismans 80      | 3D  | 0-100  | F,T    | TF     | 7    | NL  | X   | X   |
| Blankevoort 91  | 3D  | 0-90   | F,T    | TF     | 12   | NL  | TLE | X   |
| Tumer 93        | 2D  | 0-90   | F,T,P  | TF, PF | 4    | NL  | X   | X   |
| Abdel-Rahman 93 | 2D  | 0-90   | F,T    | TF     | 10   | NL  | X   | X   |
| Wilson 97       | 3D  | 0-100  | F,T    | TF     | 3    | Iso | X   | X   |
| Shelburne 97    | 2D  | 0-90   | F,T,P  | TF,PF  | 11   | NL  | X   | X   |
| Abdel-Rahman 98 | 3D  | 0-90   | F,T    | TF     | 12   | NL  | X   | X   |
| Akalan 07       | 3D  | 0-90   | F,T    | TF     | 12   | NL  | X   | X   |
| Guess 10        | 3D  | GR     | F,T,P  | TF,PF  | 10   | NL  | TLE | FE  |
| Baldwin 12      | 3D  | 0-100  | F,T,P  | TF,PF  | 6    | FE  | X   | X   |
| Kiapour 14      | 3D  | 0-50   | F,T,P  | TF,PF  | 15   | FE  | An  | FE  |
| Gerus 13        | 3D  | GR     | F,T    | TF     | CL   | NL  | X   | X   |
| Harris 16       | 3D  | 0-100  | F,T    | TF     | 14   | NL  | An  | X   |
| Ali 17          | 3D  | GR     | F,T,P  | TF,PF  | 14   | NL  | An  | X   |
| Ren 22          | 3D  | X      | F,T    | TF     | 4    | FE  | An  | FE  |

This table compiles the primary features of knee models identified by reviewing Web of Science, ScienceDirect, and PubMed articles. Each model's assumptions, parameters, and validation methods are detailed. **Abbreviations:** Dim - dimension, RoM - Range of Motion, G - geometry, Mod - model, Lig - ligament, Car - cartilage, Men- meniscus, 3D - Three-dimensional, 2D - Two-dimensional, GR - gait range, F - Femur, T - Tibia, P - Patella, TF - Tibio-Femoral interaction, PF - Patello-Femoral interaction, CL- cruciate ligaments, Def - deformable, NL - non-linear, Iso - isometric, X - property is absent, TLE - thin linear elastic on rigid foundation, An - anatomical, FE - Finite Element,

## APPENDIX B. MATLAB CODE USED IN THE STUDY

### B.1 Ligament Force generator for optimization of $\epsilon_1$

```

\footnotesize
function [F] = lig_force(K1, K2, L_fe_L, L_fe_U, e, L_low,
    L_up, inc)
% A function generates two-dimensional array used for
    visualization and optimization of
% Epsilon0 for smooth transition from toe (quadratic)
    region to linear region of
% Ligament Force. For details refer to PhD disertation by
    Kuchimov et al..
% F = lig_force(L0j) generates ligament force array versus
    ligament length
% and returns to generated array(matrix) and critical
    points, where K2(KQ in Text) is
% parabolic end and K1(KL inText) is linear beginning of
    the defined function
if nargin==3 & nargin==4;
L_fe=L_fe_L:1:L_fe_U;
L0j=L_fe/e;
L_r=L_low:inc:L_up;
L=zeros(length(L_fe),length(L_r));
F=zeros(size(L));
eps=zeros(size(L));
k=0; j=0;
for x = 1:1:length(L_fe)
    L(x,:)=(L_low):inc:(L_up);
    eps(x,:)=(L(x,:)-L0j(x))/L0j(x);
    for i=1:1:length(L)

```

```
if eps(x,i)<0;
    F(x,i)=0; k=i;
else if eps(x,i)<0.06 | eps(x,i) ==0;
    F(x,i)=K2*(L(x,i)-L0j(x))^2; j=i;
else eps(x,i)>0.06 | eps(x,i) ==0.06;
    F(x,i)=K1*(L(x,i) -(1+0.03)*L0j(x));
end
end
end
end;
end;
```

## B.2 Global Coordinate System (GCS) to Local Coordinate System (LCS) Rotation Matrix

```
function [R, alfa_beta_psi] = glob2loc
% Caution!!! Clinical rotations - flexion, abduction and
% ext rotation - are not Joint rotations angles.
% alfa = Flexion (+ve)
% beta = pi/2+Adduction (+ve right, - ve left limb)
% psi = External rot (+ve)
% They are Euler angles to elign hypothetical GCS to
% LCS by rotating frame about its Z(hGCS), X(hGCS)
% and Z(hGCS) axes.
format long
% Initialization of GCS unit axes
X=[1; 0; 0];
Y=[0; 1; 0];
Z=[0; 0; 1];
% Initialization of LCS with respect to GCS
x1=[];
y1=[];
z1=[];
% read LCS axis w.r.t. GCS
Triad = input('Please enter Triad Matrix 3X3 that defines
    local coordinate system w.r.t. GCS: ')
x1 = Triad(1,:);
y1 = Triad(2,:);
z1 = Triad(3,:);
% The first step is to find rotation angles
% They are calculated by using vectors.
% Alfa is rotation around Z1 axis to place X1 axis into XY
    plane of GCS.
```

```

x11=cross(z1,Z); % local X1 axis rotated to XY plane of
    GCS which is perpendicular to Z of GCS and Z1 local CS
x11x=dot(x11,x1); % projection of x11 (x-axis after first
    rotation around z1)on x1 (non rotated LCS x-axis)
x11y=dot(x11,y1); % projection of x11 (x-axis after first
    rotation around z1)on y1 (non rotated LCS y-axis)
alfa_ = atan2(x11y,x11x); % first rotation around z1-axis (
    LCS) in rads
alfa_deg=rad2deg(alfa_); % first rotation around z1-axis (
    LCS)in degs
% Beta is angle of rotation around LCS X1' after rotation
    alfa
% around z1 LCS x -axis => x11
z1_XY=sqrt(z1(1)^2+z1(2)^2); % projection of z1 = z-axis of
    LCS on XY plane of GCS
beta_=atan2(z1_XY,z1(3));
beta_deg=rad2deg(beta_);
% The last angle is psi. This is the angle between LCS X'
    and GCS X
% It can be found in the same way: psi = -atan2 (X'y, X'x)
    = -atan2 (-Z1x, Z1y)
psi_=-atan2(-z1(1),z1(2));
psi_deg=rad2deg(psi_);
% GCS points can be transfered to LCS (DICOM to Femur LCS
    or Tibia LCS)
a= alfa_;
b= beta_;
p= psi_;
% First column of Rotation matrix for Euler sequence 313 =
    ZXZ
R11=(cos(a)*cos(p)-sin(a)*cos(b)*sin(p));

```

```

R21=(sin(a)*cos(p)+cos(a)*cos(b)*sin(p));
R31=(sin(b)*sin(p));
%Second column of Rotation matrix
R12=(-cos(a)*sin(p)-sin(a)*cos(b)*cos(p));
R22=(-sin(a)*sin(p)+cos(a)*cos(b)*cos(p));
R23=(sin(b)*cos(p));
%Third column of Rotation matrix
R13=(sin(a)*sin(b));
R23=(-cos(a)*sin(b));
R33=(cos(b));
%Assignment of function return R -rotation matrix and
corresponding angles
R=[R11 R12 R13; R21 R22 R23; R31 R23 R33];
alfa_beta_psi = [alfa_deg; beta_deg; psi_deg];

```

### B.3 Translation Vector H (Femur to Tibia or Tibia to Femur) to Clinical Joint Shifts (q)

```
% this script takes coordinate translation H (locates tibia
    wrt femur) array as input
% calculates clinical translation q by multiplication of
    transformation matrix V and H.
% As a result body coordinate translation H converted to
    clinical translation q using
% the equation  $q = V * H$  described in the study by Grood
    and Suntay.
% Where q and H are vectors and V is transformation matrix
% Not: by default Matlab uses radians in trigonometric
    functions.
format long
% read X-component data
Hx=input('Please enter X-component data of tibial
    translation wrt femur: ')
% read clinical drawer or anterior-posterior shift +ve ant
Hy=input('Please enter Y-component data of tibial
    translation wrt femur: ')
% read clinical distraction-compression or sup-inferior
    shift +ve distraction
Hz=input('Please enter Z-component data of tibial
    translation wrt femur: ')
% read flexion angle as an array
flex = input('Please enter rotation array of tibia about
    its x-axis +ve clockwise: ')
% read adduction angle as an array
add = input('Please enter rotation array of tibia about its
    y-axis +ve counter clockwise : ')
% Angle conversion degree to radians
```

```

alfaana = deg2rad(flex);
% Angle conversion and beta = pi/2+Adduction (+ve right, -
    ve left limb)
betanaa = (pi/2+deg2rad(add));
% define 3by3by(Length of angle array) matrix
[b, ~] = size(alfaana)
VV = zeros(3,3,b);
% combine components of translation H data
h = [Hx Hy Hz]';
for i=1:1:b
    VV(:,:,i) = [1 0 0; 0 cos(alfaana(i)) -sin(alfaana(i)); -
        cos(betanaa(i)) -sin(alfaana(i))*sin(betanaa(i)) -cos(
            alfaana(i))*sin(betanaa(i))];
    qq(:,i) = VV(:,:,i)*h(:,i);
    q = [q qq];
end

```

## B.4 Parametrization of kinematics data

```
% this script takes inverse of matrix that transforms q-
    clinical translation to
% segment or body coordinate translation H using equation  $q$ 
    =  $V * H$ 
% where  $q$  and  $H$  are vectors and  $V$  is translation matrix
    described by Grood and
% Suntay. Not: by default Matlab uses radians in
    trigonometric functions.
format long
% read clinical tibial trust or medial-lateral tibial shift
    +ve lat
q1=input('Please enter medial-lateral tibial shift +ve
    lateral direction for right knee: ')
% read clinical anterior-posterior drawer or shift +ve ant
q2=input('Please enter antterior-posterior drawer +ve
    anterior direction: ')
% read joint distraction-compression +ve distraction
q3=input('Please enter joint distraction-compression +ve
    distraction direction: ')
% read flexion angle as an array
flex = input('Please enter alfa flexion angle +ve flexion
    direction : ')
% read adduction angle as an array
add = input('Please enter adduction angle (+ve right, - ve
    left limb) : ')
% Angle conversion degree to radians
alfa = deg2rad(flex);
% Angle conversion and beta = pi/2+Adduction (+ve right, -
    ve left limb)
beta = (pi/2+deg2rad(add));
```

```

% define 3by3by(Length of angle array) matrix
[b, ~] = size(alfaaa)
VV = zeros(3,3,b);
% combine q1, q2 and q3
q = [q1 q2 q3]';
for i=1:1:b
    VV(:,:,i) = [1 0 0; 0 cos(alfaaa(i)) -sin(alfaaa(i)); -
        cos(betaaaa(i)) -sin(alfaaa(i))*sin(betaaaa(i)) -cos(
            alfaaaa(i))*sin(betaaaa(i))];
    H(:,i) = inv(VV(:,:,i))*q(:,i);
    H = [H HH];
end

```

## B.5 Clinical Joint Translation to Translation Vector (Femur to Tibia or Tibia to Femur)

```
%This function has multiple task.
%1. It takes as an input kinematics data from Natural Knee
%(NK) repository. 2. It finds max and min values for
%rotational and translational displacements. 3. It trims
%spurious data which is not repeated for all flex-extension
%cycles. 4. It splines and parametrizes with respect to
%flexion-extension data. 5. It downsamples if it is
    necessary
function Kinem=datasplit(Kin)
s=input('is data array contain time at first column? Press
    Y','s' );
if s=='y'
    time = Kin(:,1);
    Kin=Kin(:,2:end);
end
lmap = [0 0 0; 1 0 0; 1 0 1; 1 1 0; 0 0 1; 0 1 0];
markk = {'+', 'o', '*', 'x', 'v', 'd', '^', 's', '>', '<'};
disp('Check for reasonable local min and max that define
    new independent axis as flexion angle');
[Peaks_PosFE, P_locs]=findpeaks(Kin(:,1));
[Peaks_NegFE, N_locs]=findpeaks(-Kin(:,1));
Peaks_NegFE = -1*Peaks_NegFE; %flipping values
figure;
plot(Kin(:,1))
hold on;
scatter(P_locs, Peaks_PosFE, 'o');
scatter(N_locs, Peaks_NegFE, 'x');
title('Flexion and Extension Peaks','FontSize',12);
xlabel('Time [1/100 s]','FontSize',12);
```

```

ylabel('Flex-extension rotation [degrees]','FontSize',12);
hold off;
MinAx = fix(max(Peaks_NegFE));
if MinAx<fix(Kin(1,1))
    MinAx=fix(Kin(1,1)+1);
end
MaxAx = fix(min(Peaks_PosFE));
Peaks_NegFE = [Kin(1,1); Peaks_NegFE]; % Add start (
    begining point)
N_locs = [1; N_locs]; % Add start index
locs=sort([P_locs; N_locs]);
intervals = diff(locs);
intervalss =sort(intervals);
num_locs=length(locs);
num_int=length(intervals);
max_int = intervalss(end);
locsi = int64(locs);
FE=MinAx:0.25:MaxAx;
NKin=nan(max_int,6,num_int);
TrimKin=nan(6,length(FE),num_int);
lmap = [0 0 0; 1 0 0; 1 0 1; 1 1 0; 0 0 1; 0 1 0];
charcol=['k'; 'r'; 'm'; 'y'; 'b'; 'g'];
markk = {'+', 'o', '*', 'x', 'v', 'd', '^', 's', '>', '<'};

for k = 1:1:6
    figure;
    for i=1:1:num_locs-1
        TrimKin(:, :, i)=spline(Kin(locsi(i)+1:locsi(i+1),1)',Kin(
            locsi(i)+1:locsi(i+1),1:6) ',FE);
        plot(FE,TrimKin(k, :, i));
    end
end

```

```

temp=mean(TrimKin(k,:,:),3);
scatter(FE,temp', 50, char(markk(1,i)), charcol(k) );
legend
title('Interpolated, Trimmed and Mean of Varum-valgum
      Rotation','FontSize',12);
xlabel('Flex-extension rotation [degrees]','FontSize',12);
ylabel('Varum-valgum Rotation','FontSize',12);
axis tight
hold off;
end

for k = 1:1:6
    figure(k);
    for i=1:1:num_locs-1
        hh=plot(Kin(locsi(i)+1:locsi(i+1),1),Kin(locsi(i)+1:
            locsi(i+1),k));
        hold on
        set(hh, 'Marker', char(markk(1,i)));
        set(hh, 'MarkerSize', 10);
        set(hh, 'LineWidth', 1);
        set(hh, 'Color', lmap(k,:));
    end
    title('Flexion and Extension Domain','FontSize',12);
    xlabel('Flex-extension rotation [degrees]','FontSize',12);
    ylabel('Clinical Translations','FontSize',12);
    axis tight
    yy=ylim;
    plot([60 60],[yy(1) yy(2)], 'k', 'LineWidth', 5);
    hold off
end

```

```

Kinem=TrimKin;
maxx=[];
minn=[];
for k = 1:1:6
    figure(k);
    maxx(k)=max(Kin(:,k));
    minn(k)=min(Kin(:,k));
    plot(downsample(time,10),downsample(Kin(:,1),10)/(maxx
        (1)-minn(1)), 'Color', 'k', 'LineWidth', 3);
    hold on;
    for i=1:1:num_locs-1
        scatter(downsample(time(locsi(i)+1:locsi(i+1)),10),
            downsample(Kin(locsi(i)+1:locsi(i+1),k),10)/((
                maxx(k)-minn(k)), 50, char(markk(1,i)),
                charcol(k));
        yy=ylim;
        line([time(locsi(i+1)) time(locsi(i+1)+2)], [yy
            (1) yy(2)] );
    end

title('Translation Data Partition', 'FontSize', 12);
xlabel('Time [s]', 'FontSize', 12);
ylabel('Rotations [degrees]', 'FontSize', 12);
hold off;
end

for k = 1:1:6
    figure(k);
    for i=1:1:num_locs-1
        hh=plot(downsample(Kin(locsi(i)+1:locsi(i+1),1),10),
            downsample(Kin(locsi(i)+1:locsi(i+1),k),10));

```

```

        hold on
        set(hh, 'Marker', char(markk(1,i)));
        set(hh, 'MarkerSize', 10);
        set(hh, 'LineWidth', 1);
        set(hh, 'Color', lmap(k,:));
    end
    title('Flexion and Extension Domain','FontSize',12);
    xlabel('Flex-extension rotation [degrees]','FontSize',12);
    ylabel('Varum-valgum','FontSize',12);
    axis tight
    yy=ylim;
    plot([60 60],[yy(1) yy(2)],'k','LineWidth', 5);
    hold off
end
for k = 1:1:6
    figure;
    for i=1:1:num_locs-1
        plot(FE,Sim(k,:,i));
        hold on;
    end
    temp=mean(Sim(k,:,:),3);
    scatter(FE,temp', 50, char(markk(1,i)), charcol(k) );
    legend
    title('Interpolated, Trimmed and Mean of Varum-valgum
        Rotation','FontSize',12);
    xlabel('Flex-extension rotation [degrees]','FontSize',12);
    ylabel('Varum-valgum Rotation','FontSize',12);
    axis tight
    hold off;
end

```

## B.6 Natural Knee (NK) and Computational Knee (CM) data loading

```

% downsampled kinematics obtained from tibia and femur
% tracking which
% represented in Hinged Femur Perpendicular Moving Tibia
% set up and
% interpolated to 0.25 degree increments of knee flexion

AAdown=downsample(AA,100);
BBdown=downsample(BB,100);

AAdownT=permute(AAdown,[3 1 2]);
BBdownT=permute(BBdown,[3 1 2]);

% RMSE calculation for downsapmled data

RMSE_down = nan(421,6);

for i =1:1:421
    for j = 1:1:6
        RMSE_down(i,j)=rmse(AAdown(i,:,j),BBdown(i,:,j));
    end
end

mean(RMSE_down(:,1))
mean(RMSE_down(:,2))
mean(RMSE_down(:,3))
mean(RMSE_down(:,4))
mean(RMSE_down(:,5))
mean(RMSE_down(:,6))

max(RMSE_down(:,1))

```

```

max(RMSE_down(:,2))
max(RMSE_down(:,3))
max(RMSE_down(:,4))
max(RMSE_down(:,5))
max(RMSE_down(:,6))

figure
hold on
for i=1:1:7
    plot(AAdownT(1, :, i), AAdownT(2:6, :, i));
end
hold off

figure
hold on
for i=1:1:7
    plot(AAdownT(1, :, i), BBdownT(2:6, :, i));
end
hold off

figure
hold on
for i=1:1:7
    plot(AAdownT(1, :, i), AAdownT(2:6, :, i));
    plot(AAdownT(1, :, i), BBdownT(2:6, :, i), 'r');
end
hold off

RMSE_down = nan(421,6);
for i = 1:1:421
    for j = 1:1:6

```

```

        RMSE_down(i,j)=rmse(AAdown(i,1:7,j),BBdown(i,1:7,j)
        );
    end
end

RMSE_down_odds = nan(421,6);
for i =1:1:421
    for j = 1:1:6
        RMSE_down_odds(i,j)=rmse(AAdown(i,1:2:6,j),BBdown(i
        ,1:2:6,j));
    end
end

RMSE_down_evens = nan(421,6);
for i =1:1:421
    for j = 1:1:6
        RMSE_down_evens(i,j)=rmse(AAdown(i,2:2:6,j),BBdown(
        i,2:2:6,j));
    end
end
end

```

## B.7 Natural Knee (NK) and Computational Knee (CM) data loading

```

%%% This script plots graphics for experimental and
    simulated kinematics.
% Data for experimental and simulated kinematics must
    present in Workspace
% and variable names must be ExpA_meannn and SimA_meannn.
    It is supposed
% kinematics which is recorded with respect to time are
    parametrized to
% flex-extension. Additionally, variables are 2D array
    containing column
% arrays period mean for each kinematics are averaged for
    of curve is considered as input
% note that workspace is loaded from any ExpVsSim mat file
% and do not forget to evaluate following:
ExpA=BBdown;
SimA=AAdown;
linee_width=3;
figure
hold on;
max_ydeg = round(max(max([ExpA_meannn(:,2:3) SimA_meannn
    (:,2:3)]))));
max_ymm = round(max(max([ExpA_meannn(:,4:6) SimA_meannn
    (:,4:6)]))));
ylim([-30 40])
for i=3:-1:1
x = ExpA_meannn(:,1)';
if i == 1
    y1 = nan(size(ExpA_meannn(:,i)'));
    y11 = nan(size(SimA_meannn(:,i)'));

```

```

        y2 = ExpA_meannn(:,i+3)';
        y22 =SimA_meannn(:,i+3)';
    else
        y1 = ExpA_meannn(:,i)';
        y2 = ExpA_meannn(:,i+3)';
        y11 =SimA_meannn(:,i)';
        y22 =SimA_meannn(:,i+3)';
    end

    [AX,H1,H2] = plotyy(x,y1,x,y2,'plot');
    [AX1,H11,H22] = plotyy(x,y11,x,y22,'plot');

    set(H1,'LineStyle','-');
    set(H2,'LineStyle','-');
    set(H1,'Color',lmap(i,:));
    set(H2,'Color',lmap(i+3,:));
    set(H1,'LineWidth',linee_width);
    set(H2,'LineWidth',linee_width);

    set(H11,'LineStyle',':')
    set(H22,'LineStyle',':')
    set(H11,'Color',lmap(i,:));
    set(H22,'Color',lmap(i+3,:));
    set(H11,'LineWidth',linee_width);
    set(H22,'LineWidth',linee_width);

end

legend('Experimental','Simulated');
set(get(AX(1),'Ylabel'),'String','Rotations in Degrees')
set(get(AX(2),'Ylabel'),'String','Translations in mm')
xlabel('Flex-extension (Degrees)','FontSize',12)

```

```

title('System Response to Passive Knee Flexion','FontSize'
    ,14)
hold off;

%% This script draws mean graph and corresponding
    deviation as errorbars

clear ME
clear SE
clear MS
clear SS

% For flexion range from 7 to 112 degrees
FEE=downsample(ExpA(:,1,1),10);
FES=downsample(SimA(:,1,1),10);
ExpAA=ExpA(:, :, :);
SimAA=SimA(:, :, :);

% For flexion range from 7 to 90 degrees
FEE=downsample(ExpA(1:333,1,1),10);
FES=downsample(SimA(1:333,1,1),10);
ExpAA=ExpA(1:333, :, :);
SimAA=SimA(1:333, :, :);

%Color map for each displacement
lmap = [
    0      0      0
    1      0      0
    1      0      1
    1      1      0
    0      0      1
    0      1      0];

%% Draws both graphs for experimental and simulated mean
    kinematics with errorbars for cycle deviation

```

```

clear ME SE MS SS
figure;
hold on;
for i=2:1:6
ME=downsample(mean(ExpAA(:, :, i), 2), 10);
SE=downsample(std(ExpAA(:, :, i), 0, 2), 10);
MS=downsample(mean(SimAA(:, :, i), 2), 10);
SS=downsample(std(SimAA(:, :, i), 0, 2), 10);
errorbar(FEE, ME, SE, 'LineWidth', 2, ...
          'Color', lmap(i, :));
errorbar(FEE, MS, SS, 'LineStyle', ':', ...
          'LineWidth', 2, ...
          'Color', lmap(i, :));
end

grid on;
legend( 'NK Abd-add Rotations', 'SK Abd-add Rotations', ...
        'NK Int-ext Rotations', 'SK Int-ext Rotations', ...
        'NK Med-lat Translations', 'SK Med-lat Translations'
        , ...
        'NK Ant-post Translations', 'SK Ant-post
        Translations', ...
        'NK Sup-inferior Translations', 'SK Sup-inferior
        Translations');
ylabel('Rotation (degrees) and Translations (mm)', 'FontSize'
       , 12)
xlabel('Flex-extension (degrees)', 'FontSize', 12)
title('Kinematics for Passive Knee Flexion', 'FontSize', 14)
hold off;

```

```

%% Draws graphs for experimental mean kinematics with
    errorbars for cycle deviation

clear ME SE MS SS
figure(1);
hold on;
for i=2:1:6
ME=downsample(mean(ExpAA(:, :, i), 2), 10);
SE=downsample(std(ExpAA(:, :, i), 0, 2), 10);
errorbar(FEE, ME, SE, 'LineWidth', 2, ...
        'Color', lmap(i, :));
end

grid on;
legend('Abd-add Rotations', 'Int-ext Rotations', 'Med-lat
    Translations', ...
        'Ant-post Translations', 'Sup-inferior Translations
        ');
ylabel('Rotation (Degrees) and Translations (mm)', 'FontSize
    ', 12)
xlabel('Flex-extension (Degrees)', 'FontSize', 12)
title('Natural Knee (NK) Experimental Kinematics', 'FontSize
    ', 14)
hold off;

%% Draws graphs for simulated mean kinematics with
    errorbars for cycle deviation

clear ME SE MS SS
figure(2);
hold on;
for i=2:1:6
MS=downsample(mean(SimAA(:, :, i), 2), 10);

```

```

SS=downsample(std(SimAA(:,: ,i),0,2),10);

errorbar(FES, MS, SS, 'LineWidth',2,...
          'Color', lmap(i,:));
end
grid on;
legend('Abd-add Rotations', 'Int-ext Rotations', 'Med-lat
       Translations', ...
       'Ant-post Translations', 'Sup-inferior Translations
       ');

ylabel('Rotation (Degrees) and Translations (mm)','FontSize
       ',12)
xlabel('Flex-extension (Degrees)','FontSize',12)
title('Computational Model (CM) Simulated Kinematics','
       FontSize',14)
hold off;

%% Draws graphs for experimental vs simulated mean
    kinematics with errorbars for cycle deviation
clear ME SE MS SS

for i=1:1:6
    figure(i);
    hold on;
    ME=downsample(mean(ExpAA(:,: ,i),2),10);
    SE=downsample(std(ExpAA(:,: ,i),0,2),10);
    MS=downsample(mean(SimAA(:,: ,i),2),10);
    SS=downsample(std(SimAA(:,: ,i),0,2),10);

    errorbar(FEE, ME, SE, 'LineWidth', 2, 'Color', 'r');

```

```

        errorbar(FEE, MS, SS, 'LineWidth', 2, 'Color', 'b');
    grid on;
    legend('Experimental', 'Simulated');
    if i<4
        ylabel('Rotation (Degrees)', 'FontSize', 12)
    else
        ylabel('Translations (mm)', 'FontSize', 12)
    end
    xlabel('Flex-extension (Degrees)', 'FontSize', 12)

    switch i
        case 1
            title('Flex-Extension Response', 'FontSize', 14)
        case 2
            title('Abd-Adduction Response', 'FontSize', 14)
        case 3
            title('Int-External Response', 'FontSize', 14)
        case 4
            title('Med-Lateral Response', 'FontSize', 14)
        case 5
            title('Ant-Posterior Response', 'FontSize', 14)
        case 6
            title('Sup-Inferior Response', 'FontSize', 14)
        end
    hold off;
end

```

## B.8 Natural Knee (NK) and Computational Knee (CM) data loading

```
% Mean and STD of 7 passive flexion motion (4 ext to flex
    and 3 flex to ext strating from knee at full extension
% knee flexed to aproximate 110 degrees and extenden back.
    Motion perfomed for 4 cycles but last extension
% was not completed and therefore it was not included)
    calculated using
% downsampled data

Mean_AAdownT_Sim=mean(AAdownT,3);
Mean_BBdownT_Exp=mean(BBdownT,3);

STD_AAdownT_Sim=std(AAdownT,1,3);
STD_BBdownT_Exp=std(BBdownT,1,3);

for k=2:1:6
    figure(k)
    hold on
    for l=1:1:7
        plot(AAdownT(1,:,l),AAdownT(k,:,l));
        plot(BBdownT(1,:,l),BBdownT(k,:,l),'r');
    end
    plot(BBdownT(1,:,l),RMSE_down(:,k),'kd');
    plot(AAdownT(1,:,l),STD_AAdownT_Sim(k,:));
    plot(BBdownT(1,:,l),STD_BBdownT_Exp(k,:), 'r');
    hold off
end
```

```

% mean of all RMSE error between experimental and simulated
    kinematics
Mean_RMSE_down = [
mean(RMSE_down(:,1))
mean(RMSE_down(:,2))
mean(RMSE_down(:,3))
mean(RMSE_down(:,4))
mean(RMSE_down(:,5))
mean(RMSE_down(:,6))];

% max of all RMSE error between experimental and simulated
    kinematics
Max_RMSE_down = [
max(RMSE_down(:,1))
max(RMSE_down(:,2))
max(RMSE_down(:,3))
max(RMSE_down(:,4))
max(RMSE_down(:,5))
max(RMSE_down(:,6))];

% range of excursion for 5 motion revealed during passive
    knee flexion
% calculations for simulated passive knee flexion
Ranges_Sim = [
range(Mean_AAdownT_Sim(1,:))
range(Mean_AAdownT_Sim(2,:))
range(Mean_AAdownT_Sim(3,:))
range(Mean_AAdownT_Sim(4,:))
range(Mean_AAdownT_Sim(5,:))
range(Mean_AAdownT_Sim(6,:))];
% calculations for experimental passive knee flexion

```

```

Ranges_Exp = [
range(Mean_BBdownT_Exp(1,:)); range(Mean_BBdownT_Exp(2,:))
range(Mean_BBdownT_Exp(3,:)); range(Mean_BBdownT_Exp(4,:))
range(Mean_BBdownT_Exp(5,:)); range(Mean_BBdownT_Exp(6,:))
];

% seperation to vectors
FE_Exp=permute(Exp(1,:,:),[2,3,1]);
FE_Sim=permute(Sim(1,:,:),[2,3,1]);
FE_SimMean=mean(FE_Sim');

VV_Exp=permute(Exp(2,:,:),[2,3,1]);
VV_Sim=permute(Sim(2,:,:),[2,3,1]);
VV_SimMean=mean(VV_Sim');

EI_Exp=permute(Exp(3,:,:),[2,3,1]);
EI_Sim=permute(Sim(3,:,:),[2,3,1]);
EI_SimMean=mean(EI_Sim');

ML_Exp=permute(Exp(4,:,:),[2,3,1]);
ML_Sim=permute(Sim(4,:,:),[2,3,1]);
ML_SimMean=mean(ML_Sim');

AP_Exp=permute(Exp(5,:,:),[2,3,1]);
AP_Sim=permute(Sim(5,:,:),[2,3,1]);
AP_SimMean=mean(AP_Sim');

SI_Exp=permute(Exp(6,:,:),[2,3,1]);
SI_Sim=permute(Sim(6,:,:),[2,3,1]);
SI_SimMean=mean(SI_Sim');

```

```

FEfit=fitlm(FE_Exp,FE_SimMean)
VVfit=fitlm(VV_Exp,VV_SimMean)
EIfit=fitlm(EI_Exp,EI_SimMean)
MLfit=fitlm(ML_Exp,ML_SimMean)
APfit=fitlm(AP_Exp,AP_SimMean)
SIfit=fitlm(SI_Exp,SI_SimMean)

figure; plot(FEfit)
figure; plot(VVfit)
figure; plot(EIfit)
figure; plot(MLfit)
figure; plot(APfit)
figure; plot(SIfit)

AASim=mean(AA,2); %takes mean along 2 dimension of all 7
    samples
AAASim=permute(AASim,[1 3 2]); % converts 3D to 2D matrix

BBExp=mean(BB,2); %takes mean along 2 dimension of all 7
    samples
BBBExp=permute(BBExp,[1 3 2]); % converts 3D to 2D matrix

%R squared evaluation for means of the Experimental and
    Simulated data

R2_ExpVSsimMeans = nan(1,6);

for i =1:1:6
    R2_ExpVSsimMeans(i)=rsquare(BBBExp(:,i),AAASim(:,i));

```

```

end

FEfit=fitlm(BBBExp(:,1),AASim(:, :, 1));
VVfit=fitlm(BBBExp(:,2),AASim(:, :, 2));
IEfit=fitlm(BBBExp(:,3),AASim(:, :, 3));
MLfit=fitlm(BBBExp(:,4),AASim(:, :, 4));
APfit=fitlm(BBBExp(:,5),AASim(:, :, 5));
SIfit=fitlm(BBBExp(:,6),AASim(:, :, 6));


Exp=TrimKin;
ExpA=permute(Exp,[2 3 1]);
ExpA_mean=mean(ExpA,2);
ExpA_meann=permute(ExpA_mean,[1 3 2]);


Sim=TrimKin;
SimA=permute(Sim,[2 3 1]);
SimA_mean=mean(SimA,2);
SimA_meann=permute(SimA_mean,[1 3 2]);


%% Linear Regression applied Simulated (Model response X)
    based the Experimental (Measured Y)
%Note that experimental and simulated arrays must be in the
    workspace as
%ExpA_meannn and SimA_meannn


FEfit=fitlm(ExpA_meannn(:,1),SimA_meannn(:,1))
VVfit=fitlm(ExpA_meannn(:,2),SimA_meannn(:,2))
IEfit=fitlm(ExpA_meannn(:,3),SimA_meannn(:,3))
MLfit=fitlm(ExpA_meannn(:,4),SimA_meannn(:,4))
APfit=fitlm(ExpA_meannn(:,5),SimA_meannn(:,5))
SIfit=fitlm(ExpA_meannn(:,6),SimA_meannn(:,6))

```

```
figure; plot(FEfit)
figure; plot(VVfit)
figure; plot(IEfit)
figure; plot(MLfit)
figure; plot(APfit)
figure; plot(SIfit)
```

```
FEfit=fitlm(ExpA_meannn(:,1),SimA_meannn(:,1))
VVfit=fitlm(ExpA_meannn(:,2),SimA_meannn(:,2))
IEfit=fitlm(ExpA_meannn(:,3),SimA_meannn(:,3))
MLfit=fitlm(ExpA_meannn(:,4),SimA_meannn(:,4))
APfit=fitlm(ExpA_meannn(:,5),SimA_meannn(:,5))
SIfit=fitlm(ExpA_meannn(:,6),SimA_meannn(:,6))
```

```
figure; plot(FEfit)
figure; plot(VVfit)
figure; plot(IEfit)
figure; plot(MLfit)
figure; plot(APfit)
figure; plot(SIfit)
```

```
% Linear Regression applied for the range of 7-90 flexion
    degrees Simulated
%(Model response X) based the Experimental (Measured Y)
%Note that experimental and simulated arrays just trancated
    not splined for that range
% Data must exist in the workspace as ExpA_meannn and
    SimA_meannn
```

```
FEfit1=corrcoef(ExpA_meannn(1:333,1),SimA_meannn(1:333,1))
```

```

VVfit1=corrcoef(ExpA_meannn(1:333,2),SimA_meannn(1:333,2))
IEfit1=corrcoef(ExpA_meannn(1:333,3),SimA_meannn(1:333,3))
MLfit1=corrcoef(ExpA_meannn(1:333,4),SimA_meannn(1:333,4))
APfit1=corrcoef(ExpA_meannn(1:333,5),SimA_meannn(1:333,5))
SIfit1=corrcoef(ExpA_meannn(1:333,6),SimA_meannn(1:333,6))

figure; plot(FEfit1)
figure; plot(VVfit1)
figure; plot(IEfit1)
figure; plot(MLfit1)
figure; plot(APfit1)
figure; plot(SIfit1)

lmap = [0 0 0; 1 0 0; 1 0 1; 1 1 0; 0 0 1; 0 1 0];
markk = {'+', 'o', '*', 'x', 'v', 'd', '^', 's', '>', '<'};
YYlabel = {'Flex-extension Rotation', 'Varum-valgum
           Rotation', 'Inter-external Rotation', 'Medio-lateral
           Translation', 'Anterio-posterior Translation', 'Super-
           inferior Translation'};
for i=1:1:6
    figure;
    hh=plot(ExpA_meannn(:,1),ExpA_meannn(:,i));
    hold on;
    hhh=plot(ExpA_meannn(:,1),SimA_meannn(:,i));
    legend('Experimental', 'Simulated');
    set(hhh, 'Marker', char(markk(1,i)));
    set(hh, 'MarkerSize', 10);
    set(hh, 'LineWidth', 10);
    set(hh, 'Color', lmap(i,:));
    title('Mean of Experimental and Model Responce', '
           FontSize',12);

```

```

xlabel('Flex-extension rotation [degrees]','FontSize'
      ,12);
ylabel(YYlabel(i),'FontSize',12);
axis tight
hold off;
end

marker_size = 5;
line_with = 5;

figure;
hold on;
for i=2:1:6
    hh=plot(ExpA_meannn(:,1),ExpA_meannn(:,i));
    set(hh,'LineStyle',':')
    set(hh,'LineWidth', line_with);
    set(hh,'Color', lmap(i,:));
    hhh=plot(ExpA_meannn(:,1),SimA_meannn(:,i));
    set(hhh,'LineStyle','-')
    set(hhh,'LineWidth', line_with);
    set(hhh,'Color', lmap(i,:));
end
title('Mean of Experimental and Model Responce','
      FontSize',24);
xlabel('Flex-extension rotation [degrees]','FontSize'
      ,18);
ylabel('Angular and Linear Translations','FontSize',18)
;
legend('Exp Var-Val', 'Sim Var-Val','Exp Int-Ext', 'Sim
      Int-Ext','Exp Med-Lat', 'Sim Med-Lat','Exp Ant-Post
      ', 'Sim Ant-Post', 'Exp Sup-Inf', 'Sim Sup-Inf');

```

```

        axis tight
        hold off;

marker_size = 5;
line_with = 2;

figure;
hold on;
for i=2:1:6
    [AX, H1, H2] = plotyy(ExpA_meannn(:,1),ExpA_meannn(:,i)
        ,ExpA_meannn(:,1),SimA_meannn(:,i),'plot');
    set(H1, 'Marker', 'x');
    set(H1, 'MarkerSize', marker_size);
    set(H1, 'LineWidth', line_with);
    set(H1, 'Color', lmap(i,:));
    set(H2, 'Marker', 'o');
    set(H2, 'MarkerSize', marker_size);
    set(H2, 'LineWidth', line_with);
    set(H2, 'Color', lmap(i,:));
    legend('Experimental', 'Simulated');
end

title('Mean of Experimental and Model Responce','
    FontSize',12);
xlabel('Flex-extension rotation [degrees]','FontSize'
    ,12);
ylabel(YYlabel(i),'FontSize',12);
axis tight
hold off;

marker_size = 5;
line_with = 2;

```

```

figure;
[AX, H1, H2] = plotyy(ExpA_meannn(:,1)',ExpA_meannn
   (:,2)',ExpA_meannn(:,1)',SimA_meannn(:,2) ','plot');
title('Mean of Experimental and Model Responce','
    FontSize',12);
xlabel('Flex-extension rotation [degrees]','FontSize'
    ,12);
ylabel(YYlabel(i),'FontSize',12);
axis tight

```

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