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Effect of Patellar Component Thickness on Patellar Kinematics and Patellofemoral Joint Function Following Total Knee Replacement

Xin Xie
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EFFECT OF PATELLAR COMPONENT THICKNESS ON PATELLAR KINEMATICS AND PATELLOFEMORAL JOINT FUNCTION FOLLOWING TOTAL KNEE REPLACEMENT

A Dissertation
Presented to
the Graduate School of
Clemson University

In Partial Fulfillment
of the Requirements for the Degree
Doctor of Philosophy
Bioengineering

by
Xin Xie
August 2014

Accepted by:
Martine LaBerge, PhD, Committee Chair
John DesJardins, PhD, Co-Committee Chair
Hai Yao, PhD
Lonny Thompson, PhD
Frank Voss, MD
ABSTRACT

Nowadays, instances of anterior knee pain and patellar fracture are more and more widely considered as significant complications following total knee replacement (TKR). The inability to freely flex/extend the knee after surgery has a critical influence on patients’ daily activities, which is one of the most common mechanical indications for TKR revisions. An overtly thick or thin patellar component selected intraoperatively in TKR surgery may cause postoperative complications, including excessive quadriceps tendon force, patellofemoral joint reaction force, and joint pain in TKR patients during knee joint motion.

The objective of the current study was to evaluate the effect of varying patellar component thickness on patellar kinematics and patellofemoral joint function following TKR. Majorly, this project consisted of three parts: (a) combined computational and experimental exploration of knee joint alignment and tibiofemoral joint mechanics; (b) computational investigation of patellofemoral joint contact pressure and quadriceps tendon force; (c) comprehensive assessment of patellar kinematics and patellofemoral joint function utilizing in-vitro cadaveric testing.

Within the first section, a 3D computational finite element (FE) model was developed based on the Stanmore knee joint simulator outputs and experimentally validated to optimize the tibiofemoral rotational alignment for appropriate patellar tracking and reduced tibiofemoral contact pressure.

Secondly, by further enhancing the computational simulation, a 3D knee joint model with was established using realistic anatomical geometry, material property and
intraoperatively kinematic/kinetic boundary conditions. A commercially available TKR system was inserted in the FEA model. Quadriceps tendon force and patellofemoral contact pressure at multiple patellar thickness levels were computationally predicted and thus their variation pattern along with patellar thickness change was derived.

Thirdly, a cadaver study was conducted to comprehensively assess the biomechanical influence of patellar thickness change on patellar kinematics and patellofemoral joint function of three different thickness levels using a customized UHMWPE pressure sensor with advanced mapping capability.

Overall, the combined computational and experimental studies revealed that increase of the patellar thickness contributes to extra patellar lateral tilt, lateral shift and patellar flexion during knee joint motion. Also, the thicker patella leads to lower quadriceps tendon force at lower knee flexion angles, but higher force magnitudes at deep flexion angles. Thinner patella is believed to be associated with excessive quadriceps tendon force during knee extension, but the thickness reduction benefits patellofemoral joint function by decreasing peak patellofemoral contact pressure.

This study focused on the patellofemoral joint as a major cause of TKR failure. A direct outcome of this study was the development of an intraoperative predictive reference for practitioners in selecting the appropriate patellar component thickness to mitigate clinical failure. Novel methods and technologies utilized in the current study can be expected to advance the state of the art in real-time contact pressure measurement for congruous surfaces, advanced modeling of the complex knee bearing system as well
biological tissues, and in-vitro testing solution for investigation of knee joint biomechanics.
DEDICATION

This work is dedicated to the most important people in my life.

God: thank you for your unreserved love and thank you for the wisdom you provide me to accomplish this work.

My parents: thank you for your unconditional support with my studies. Thank you for your understanding and encouragement.

My wife: thank you for your endurance and your constant love. Because of you, my life at Clemson is full of laughs!

My church friends: I am so thankful for having you as my sisters and brothers. Thank you for your affirmation and prayers.
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I want to extend my gratitude to Dr. Andrew Clark for his interest and investment in my projects. His expertise enabled me to accomplish my work. Thanks especially to Dr. Julia Lee and Mr. Dennis Chou for assisting my experimental testing.

Throughout my time at Clemson, I have been benefited from the teaching of several great professors. My sincere thanks go to Dr. Mica Grujicic, Dr. Nicole Coutris, Dr. Delphine Dean and Dr. Melinda Harman.

Lastly, I want to thank my fellow graduate students and friends for their kindness and understanding. They truly enhanced my experience at Clemson.
# TABLE OF CONTENTS

<table>
<thead>
<tr>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>TITLE PAGE</td>
</tr>
<tr>
<td>ABSTRACT</td>
</tr>
<tr>
<td>DEDICATION</td>
</tr>
<tr>
<td>ACKNOWLEDGMENTS</td>
</tr>
<tr>
<td>LIST OF TABLES</td>
</tr>
<tr>
<td>LIST OF FIGURES</td>
</tr>
<tr>
<td>CHAPTER</td>
</tr>
<tr>
<td>1. INTRODUCTION</td>
</tr>
<tr>
<td>1.1 TKR Overview</td>
</tr>
<tr>
<td>1.2 Patella and Quadriceps Tendon --- Overview</td>
</tr>
<tr>
<td>1.3 Knee Anatomy</td>
</tr>
<tr>
<td>1.4 Knee Joint Kinematics and Kinetics</td>
</tr>
<tr>
<td>2. RESEARCH OVERVIEW</td>
</tr>
<tr>
<td>2.1 Overview</td>
</tr>
<tr>
<td>2.2 Specific Aims</td>
</tr>
<tr>
<td>3. EFFECT OF ROTATIONAL PROSTHETIC ALIGNMENT VARIATION ON THE TIBIOFEMORAL CONTACT PRESSURE AND JOINT KINEMATICS IN TOTAL KNEE REPLACEMENT</td>
</tr>
<tr>
<td>3.1 Introduction</td>
</tr>
<tr>
<td>3.2 Methods</td>
</tr>
<tr>
<td>3.3 Results</td>
</tr>
<tr>
<td>3.4 Discussion</td>
</tr>
<tr>
<td>3.5 Conclusions</td>
</tr>
<tr>
<td>3.6 Acknowledgement</td>
</tr>
</tbody>
</table>
# Table of Contents (Continued)

<table>
<thead>
<tr>
<th>Section</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>4. <strong>INTRODUCTION OF EFFECT OF PATELLAR COMPONENT THICKNESS ON PATELLAR KINEMATICS AND PATELLOFEMORAL JOINT MECHANICS FOLLOWING TOTAL KNEE REPLACEMENT</strong></td>
<td>71</td>
</tr>
<tr>
<td>5. <strong>COMPUTATIONAL INVESTIGATION ON EFFECT OF PATELLAR COMPONENT THICKNESS ON QUADRICEPS TENDON FORCE AND PATELLOFEMORAL CONTACT PRESSURE</strong></td>
<td>76</td>
</tr>
<tr>
<td>5.1 Methods</td>
<td>76</td>
</tr>
<tr>
<td>5.2 Results</td>
<td>90</td>
</tr>
<tr>
<td>5.3 Discussion and Conclusions.</td>
<td>93</td>
</tr>
<tr>
<td>6. <strong>DEVELOPMENT OF A NOVEL UHMWPE PATELLAR SENSOR FOR MEASURING DYNAMIC PATHELLOFEMORAL CONTACT PRESSURE</strong></td>
<td>95</td>
</tr>
<tr>
<td>6.1 Introduction</td>
<td>95</td>
</tr>
<tr>
<td>6.2 Methods</td>
<td>96</td>
</tr>
<tr>
<td>6.3 Results</td>
<td>101</td>
</tr>
<tr>
<td>6.4 Discussion and Conclusions.</td>
<td>102</td>
</tr>
<tr>
<td>7. <strong>EFFECT OF PATELLAR COMPONENT THICKNESS ON PATELLAR KINEMATICS AND PATELLOFEMORAL JOINT MECHANICS FOLLOWING TOTAL KNEE REPLACEMENT---A CADAVER STUDY</strong></td>
<td>105</td>
</tr>
<tr>
<td>7.1 Methods</td>
<td>105</td>
</tr>
<tr>
<td>7.2 Results</td>
<td>112</td>
</tr>
<tr>
<td>7.3 Discussion</td>
<td>125</td>
</tr>
<tr>
<td>7.4 Conclusions</td>
<td>132</td>
</tr>
<tr>
<td>7.5 Acknowledgement</td>
<td>133</td>
</tr>
<tr>
<td>8. <strong>CONCLUSIONS</strong></td>
<td>134</td>
</tr>
<tr>
<td>9. <strong>RECOMMENDATIONS FOR FUTURE WORK</strong></td>
<td>136</td>
</tr>
</tbody>
</table>
Table of Contents (Continued)

<table>
<thead>
<tr>
<th>Reference</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>REFERENCE</td>
<td>138</td>
</tr>
</tbody>
</table>
## LIST OF TABLES

<table>
<thead>
<tr>
<th>Table</th>
<th>Description</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>3.1</td>
<td>Material parameters for defining the CoCr and Ti alloys</td>
<td>41</td>
</tr>
<tr>
<td>3.2</td>
<td>Cases selected for further FE simulation</td>
<td>44</td>
</tr>
<tr>
<td>Figure</td>
<td>Description</td>
<td>Page</td>
</tr>
<tr>
<td>--------</td>
<td>-------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------</td>
<td>------</td>
</tr>
<tr>
<td>2.1</td>
<td>Schematic flow chart of the current project</td>
<td>32</td>
</tr>
<tr>
<td>3.1</td>
<td>Waveform inputs of femoral flexion angle (a), tibial internal (+)-external (-) torque (b) and anterior (+)-posterior (-) translational force (c), and tibial axial compressive loading (d) for the Instron-Stanmore knee simulator</td>
<td>38</td>
</tr>
<tr>
<td>3.2</td>
<td>CAD geometry of the phenolic block for fixing the TKR femoral component</td>
<td>39</td>
</tr>
<tr>
<td>3.3</td>
<td>(a) Finite element model of the Instron-Stanmore simulator testing components and (b) side view of the femoral bracket and top view of the tibial assembly during the corresponding experimental testing</td>
<td>42</td>
</tr>
<tr>
<td>3.4</td>
<td>Representative modeling cases for varied alignment DOE study (ExR6, InR6, TL7, Val5 and Var5)</td>
<td>45</td>
</tr>
<tr>
<td>3.5</td>
<td>Comparative curves of tibiofemoral translational (a) and rotational (b) motion exported from experimental knee simulator and the standard FE model</td>
<td>47</td>
</tr>
<tr>
<td>3.6</td>
<td>(a) Variation of peak tibiofemoral contact pressure within lateral and medial condyles and (b) total contact area during the gait cycle</td>
<td>48</td>
</tr>
<tr>
<td>3.7</td>
<td>Tibiofemoral contact pressure contours at (a) 5%, (b) 15%, (c) 35%, (d) 40%, (e) 45% and (f) 55% of the gait cycle, presenting local maximal magnitudes of contact pressure and contact area</td>
<td>50</td>
</tr>
<tr>
<td>3.8</td>
<td>Comparative case outputs of the tibial internal-external rotation variation during the gait cycle with different postoperative TKR alignment: (a) TKR aligned with additional I/E angle; (b) TKR aligned with additional tibial slope; (c) TKR aligned with additional V/V angle</td>
<td>53</td>
</tr>
</tbody>
</table>
List of Figures (Continued)

<table>
<thead>
<tr>
<th>Figure</th>
<th>Description</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>3.9</td>
<td>Comparative case outputs of the tibial anterior-posterior translation variation during the gait cycle with different postoperative TKR alignment: (a) TKR aligned with additional I/E angle; (b) TKR aligned with additional tibial slope; (c) TKR aligned with additional V/V angle</td>
<td>56</td>
</tr>
<tr>
<td>3.10</td>
<td>Comparative case outputs of the peak tibiofemoral contact pressure variation during the gait cycle with different postoperative TKR alignment: (a)-(b) TKR aligned with additional I/E angle; (c) TKR aligned with additional tibial slope; (d)-(e) TKR aligned with additional V/V angle</td>
<td>59</td>
</tr>
<tr>
<td>3.11</td>
<td>Response surface obtained on the basis of post-processing DOE study and existing case study results at tibial posterior slope of (a) 1°, (b) 3°, (c) 5° and (d) 7°, indicating the influence of all 3 major TKR rotational alignment parameters on the maximum tibiofemoral contact pressure magnitude</td>
<td>62</td>
</tr>
<tr>
<td>5.1</td>
<td>Axial MRI scan of human thighs obtained from public database</td>
<td>77</td>
</tr>
<tr>
<td>5.2</td>
<td>(a) Surgical resection of the femur, (b) CAD geometry of the scanned TKR device</td>
<td>79</td>
</tr>
<tr>
<td>5.3</td>
<td>3D representation of the FE geometrical model</td>
<td>80</td>
</tr>
<tr>
<td>5.4</td>
<td>Schematic layouts of reflective markers and identifiable positioning points in both captured views</td>
<td>88</td>
</tr>
<tr>
<td>5.5</td>
<td>Marker points from the Instron-Stanmore simulator were placed in the FE model for specification of femoral and tibial rigid body motions</td>
<td>90</td>
</tr>
<tr>
<td>5.6</td>
<td>Morphed patellar component models at different thickness levels: (a) -1 mm, (b) neutral and (c) +2 mm</td>
<td>91</td>
</tr>
<tr>
<td>Figure</td>
<td>Description</td>
<td>Page</td>
</tr>
<tr>
<td>--------</td>
<td>-------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------</td>
<td>------</td>
</tr>
<tr>
<td>5.7</td>
<td>Variation of (a) quadriceps tendon force and (b) peak patellofemoral contact pressure during knee joint flexion from 0°-60°</td>
<td>92</td>
</tr>
<tr>
<td>5.8</td>
<td>(a) Computational representation of the knee joint at 60° flexion and (b) Quadriceps tendon reached the maximum deformation at 60° knee flexion</td>
<td>93</td>
</tr>
<tr>
<td>5.9</td>
<td>Patellofemoral contact pressure distribution at (a) 5° and (b) 60° knee flexion</td>
<td>93</td>
</tr>
<tr>
<td>6.1</td>
<td>(a) 3D view of block with enlarged grid pattern to illustrate how blocks were fabricated; (b) Top view of the machined and instrumented prototype of the UHMWPE patellar sensor</td>
<td>98</td>
</tr>
<tr>
<td>6.2</td>
<td>(a) The Tekscan K-scan TM system used in current study; (b) Customized stainless-steel indenter utilized in patellar sensor calibration; (c) Experimental setting for comparative sensor validation testing</td>
<td>100</td>
</tr>
<tr>
<td>6.3</td>
<td>Peak pressure-variation curves obtained by both Tekscan K-scan™ system and UHMWPE patellar sensor</td>
<td>102</td>
</tr>
<tr>
<td>6.4</td>
<td>Pressure distribution at 3 real-time loading levels presented by Tekscan (left) and UHMWPE patellar (right) sensors</td>
<td>103</td>
</tr>
<tr>
<td>7.1</td>
<td>(a) Dissected patient-specific cadaver specimen used in the current study and (b) customized tendinous tissue clamp.</td>
<td>106</td>
</tr>
<tr>
<td>7.2</td>
<td>Experimental setup for the patient-specific cadaver specimen: (a) Polaris motion tracking tools were utilized for capturing rigid body motion of 1- femur, 2- tibia and 3- patella; (b) the commercial TKR device was inserted into the specimen; (c) tendinous structures of quadriceps muscles, VM, VI/RF and VL were arranged as prescribed by Sakai et al.</td>
<td>108</td>
</tr>
</tbody>
</table>
List of Figures (Continued)

<table>
<thead>
<tr>
<th>Figure</th>
<th>Description</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>7.3</td>
<td>(a) The customized UHMWPE patellar sensor employed in the current study with the same geometrical shape as patellar component; (b) CAD model of the specific fixture attached to patella to fit the pressure sensor; (c) The patellar pressure sensor inserted into the cadaver specimen with PE shims for adjusting thickness levels</td>
<td>109</td>
</tr>
<tr>
<td>7.4</td>
<td>Patellar motion identified based upon combined femoral and patellar body fixed axes</td>
<td>113</td>
</tr>
<tr>
<td>7.5</td>
<td>From the optical motion capturing data, processed intraoperative patellar kinematics during the experimental knee joint movement at 3 patellar component thickness levels, including (a) patellar medial-lateral tilt, (b) patellar medial-lateral rotation, (c) patellar medial-lateral shift and (d) patellar flexion</td>
<td>115</td>
</tr>
<tr>
<td>7.6</td>
<td>From the optical motion capturing data, processed intraoperative patellar kinematics during the experimental knee joint movement at 3 patellar component thickness levels with lateral retinacular release (LRR) procedure performed, including (a) patellar medial-lateral tilt, (b) patellar medial-lateral rotation, (c) patellar medial-lateral shift and (d) patellar flexion</td>
<td>118</td>
</tr>
<tr>
<td>7.7</td>
<td>Variation of quadriceps tendon force during the experimental knee joint motion measured by load cell at different patellar thickness levels for (a) the normal case and (b) case with LRR procedure performed</td>
<td>121</td>
</tr>
<tr>
<td>7.8</td>
<td>Variation of peak patellofemoral contact pressure value across patellar articular surface during the experimental knee joint motion measured by the customized pressure sensor at different patellar thickness levels for (a) the normal case and (b) case with LRR procedure</td>
<td>123</td>
</tr>
</tbody>
</table>
List of Figures (Continued)

<table>
<thead>
<tr>
<th>Figure</th>
<th>Description</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>7.9</td>
<td>Real-time patellofemoral contact pressure distribution contours exported from the customized pressure sensor for (a) thinner patella case at 80°, (b) the neutral case at 50° and (c) the thicker patella case at 60°; and after LRR, (d) thinner and (e) neutral patella cases at 60°, and (f) the thicker patella case at 70° flexion.</td>
<td>125</td>
</tr>
</tbody>
</table>
CHAPTER 1  INTRODUCTION

1.1 TKR Overview

As one of the most significant orthopaedic procedures performed nowadays, total knee replacement (TKR), has gained more and more attention. Statistical data have shown that over 500,000 TKR surgeries are performed per year in the United States, and it is estimated that this number will soon increase to over 3.48 million per year because of the aging baby-boomers, more common obesity problem and the trend of extension to both younger and older patients [1]. Also, total joint replacement surgery is allowed in several cases for elderly patients with co-morbid conditions to benefit their lives [2]. Due to various reasons, increasing research interests and investment have been made regarding the implant longevity, failure mechanisms, biocompatibility and design optimization, which are some major concerns associated with TKR.

The purpose of the TKR surgery is to improve functional status, relieve pain and result in relatively low perioperative morbidity for the patients’ knees [3]. Currently, total knee arthroplasty/replacement is the most effective solution for patients with severe pain in knee, osteoarthritis, or rheumatoid arthritis [4]. Modern TKR devices majorly contain four parts: femoral component, tibial component, tibial insert (plastic spacer) and patellar component. These components are anchored to the associated bone through press-fit/bone ingrowth or bone cement. Two major articulations are observed: tibiofemoral and patella-femoral articulations, both of which are of the contact relation between metal and polyethylene. Specifically, CoCrMo and Titanium alloys are widely-applied for femoral
and tibial components, respectively. CoCrMo alloy has high stiffness modulus for the weight-bearing condition and superior property of corrosion resistance, while Titanium alloy is known as an ideal biocompatible material, which can effectively reduce the effect of stress shielding [5].

The tibial insert is made of ultra-high molecular weight polyethylene (UHMWPE) and serves as a weight bearing component. The tibial insert can absorb the energy transmitted from the femoral component, including the kinetic energy due to the relative femur-tibial motion and thermal energy generated by friction at the surface. Even though UHMWPE is widely accepted by orthopedic surgeons, its susceptibility to wear continues to be the long-term limiting factor in the life of these implants [6, 7, 8]. For the patellar component, which will be deeply discussed in the present work, it is a button-shaped part made of UHMWPE or metal which is placed at the backside against the surgically resurfaced native patella. The fitting relationship between the implant trochlea groove and patellar component resumes the fundamental tribological condition of the patellofemoral joint.

Different from the fracture fixation device, which is load sharing device, TKR device serves as load bearing device due to the particular function of knee joint, for example, as stated in literature, peak magnitude of the knee joint force remained in the range of 1.8~2.5 body weight during regular comfortable walking [9]. The tibial insert composes the weight bearing surface, and there are two types of tibial insert design regarding the bearing surfaces: fixed bearing, and mobile bearing in which certain degree of freedom of tibial insert is allowed with respect to the tibial base-plate. Previous
research found that the mobile-bearing design can reduce the rate of linear wear for the mobile high-conforming implants through knee simulator and retrieval studies [10, 11]. Also, previous finite element analyses have concluded that less polyethylene contact stresses for the mobile bearing surface design for tested pattern of motion [12, 13]. However, in-vivo kinematics studies did not show consistent conclusions with the aforementioned findings [14]. In addition, backside wear of the tibial insert triggered by relative motion between tibial insert and the tibial base-plate and potential dislocation could also cause failure of the implant.

The failure mechanisms of TKR majorly include infection, stress-shielding, peri-prosthetic osteonecrosis or fracture, implant instability, polyethylene wear-through or fracture, allergy to components and wear/osteolysis. It is also recognized that the type of implant, age and gender of the patient, diagnosis, type of fixation, and design of the patellar component are all factors affecting the TKR failure [15]. Particularly, Rand et al. concluded that the survivorship at ten years was 92% for prostheses fixed with cement compared with 61% for those fixed without cement [15], indicating the influence of the different fixation methods on the potential failure of the TKR device, specifically loosening or dislocation. In addition, it is also noted by them that survivorship at ten years was 92% for nonmodular metal-backed tibial components, 90% for modular metal-backed tibial components, and 97% for all-polyethylene tibial components [15], indicating that different designs affect the failure of the implant. Among all the risk factors, wear resulted from the implant loosening and induced osteolysis are considered to be the most common cause of implant failure. Theoretically, given that ideal
conditions are satisfied, which includes appropriate combination of patient gender, fixation type, implant design, etc., the survivorship could still hardly reach 100%. Regularly, most TKR devices fail after 10-20 years since first implantation. Hence, the ultimate goal of TKR related research is to elongate the longevity of the device, as well as to reduce unnecessary revisions for patients.

1.2 Patella and Quadriceps Tendon --- Overview

As one of the bony components in a knee joint, the patella is embedded within the quadriceps tendon, and further identified as the largest sesamoid bone (bone which is embedded within a tendon) in the human body. Patella is in the shape of an inverted triangle with apex directed inferior [16]. The patellar anterior surface is connected to the quadriceps tendon at the rough superior border and V-shaped point, respectively. The quadriceps tendon, connecting quadriceps femoris muscles to the patella and extending from the patellar to tibial tuberosity, serves as a part of extensor mechanism which composes six more muscles (the rectus femoris, the vastus intermedius, the vastus lateralis, the vastus medialis (longus), the vastus medialis (obliquus) and the articularis genu) [17].

Typically, the knee joint comprises two sub-articulation structures: the tibiofemoral (TF) joint and the patellofemoral (PF) joint. The patellofemoral joint, majorly consists of patella and femoral trochlea, is critical to the overall knee stability primarily through its role in the extensor mechanism. Patella exerts additional translational displacement onto the quadriceps tendon away from the tibiofemoral contact point during knee joint motion, therefore increasing the mechanical advantage of the
extensor mechanism by transmitting the extensor force across the knee at a greater distance from the axis of knee flexion, and increasing the moment arm of the extensor muscle contraction force [18, 19]. On the other hand, the quadriceps tension rises as the knee flexes, due to the fact that line of action of body weight moves behind the knee, which requires greater quadriceps tension to maintain overall equilibrium [20]. This is consistent with conclusion drawn by Argenson et al., who claimed that the quadriceps muscle force increases as the knee experiences higher knee flexion [21].

In general, the patella serves as a combination of linkage and pulley depending on its biomechanical function. It is also essential in centralizing the divergent forces coming from the divisions of the quadriceps muscle groups and transmitting tension around the femur from the quadriceps tendon, in a frictionless way, to the tibial tuberosity. Moreover, the patella is considered as a bony shield for the trochlea and the distal femoral condyles with the knee in flexion [19].

1.3 Knee Anatomy

1.3.1 Overview

The human knee is the junction among the femur, tibia and patella. It is a synovial joint, meaning the opposing surfaces (contacting surfaces) are covered by the hyaline articular cartilage and are enclosed in a joint cavity which contains a highly viscous synovial fluid (tribological lubricant and nutrient transportation media) between the inner synovial membrane and the outer fibrous capsule. Additional fibrous bands both outside the joint capsule and inside the cavity are called ligaments. The articular cavity is divided into two parts by the intraarticular fibrocartilage menisci [22].
As mentioned previously, two articular structures are involved in the knee, the patellofemoral joint and tibiofemoral joint. For the tibiofemoral joint, it consists of the medial and lateral condyle of the femur, each of which has a distinct shape that corresponds to the geometrical outline of the tibial plateau, which is created at the proximal end of the tibia with medial and lateral sections divided by the tibial spine. Besides division of the joint space, menisci deepens the contour of these plateaus, providing a good “seat” for the corresponding femoral condyles [17], as well as cushioning to counterbalance the collision effect onto the tibial plateau and absorb kinematic energy. Also, due to fact that the lateral femoral condyle and lateral tibial plateau are both somewhat convex, to maintain articular surface fit, the added depth by menisci is critically important [17].

Regarding the patellofemoral joint, it consists of the sulcus of the femur, or trochlear groove, and the patellar articular surface. The lateral condyle of the femur is higher than the medial and helps in preventing lateral subluxation of the patella [19]. Patellofemoral joint is closely associated with the extensor mechanism because the patella plays significant role in reducing the quadriceps tendon force by increasing the lever arm (mechanical advantage) of the extensor muscle during the knee flexion.

Because of the knee’s dual function of load transmission and motion, its ligamentous and muscular arrangement is of utmost importance in maintaining the stability necessary and restrain movement to prevent injury. In particular, the knee is stabilized muscarily on the anterior side by the quadriceps, on the medial side by the sartorius and gracilis, on the lateral side by the tensor fasciae latae, and on the posterior
side by the hamstrings from above and gastrocnemius from below \[22\]. In addition, passive ligaments are recruited for knee stabilization, including the medical collateral ligament, lateral collateral ligament, anterior cruciate ligament, posterior cruciate ligament, medial patellofemoral ligament, etc., which will be discussed later.

The aforementioned two articular structures are enclosed within the outer fibrous layer and inner synovial membrane. The capsule is attached to the femur posteriorly around the articular margins of the condyles and below the intercondylar fossa. In addition, to reinforce stability of the knee capsule, tendons and ligaments are recruited laterally and medially.

1.3.2 Bones

The knee can be simply treated as a modified hinge joint that allows flexion and rotation, but it has more complex loading conditions and motion types, meaning more than 1 degree of freedom are available. The bony architecture of the femur, tibia and patella contributes to the structural frame and stability of the knee joint, along with static and dynamic restraints of the ligaments, capsule, and musculature crossing the joint. The architecture of the bones dictates the allowed motion of the joint \[18\].

The tibiofemoral joint is the largest joint structure in the human body, and is comprised of two condyloid articulations: the medial and lateral femoral condyles. They articulate with corresponding tibial plateaus. The most critical purpose of the medial and lateral menisci is to enhance the geometrical conformity of the joint contact, and to reduce the contact stress \[18, 22\].
In the sagittal profile, the articular surface of the femoral condyle is in spiral-shaped or cam-shaped with respect to the tibiofemoral joint flexion axis, and the curvature of the condyle increases from its anterior to posterior aspect. The medial condyle has a larger radius of curvature than the lateral in both anterior-posterior and proximal-distal directions, contributing to the varus-valgus and anterior-posterior alignment of the knee, and extends distal to the lateral on the anterior-posterior projection, while the lateral condyle extends anterior to the medial side on the lateral projection [18]. In sagittal direction, the lateral condyle is 1-2 cm shorter than the medial condyle that has nearly constant width in the transverse section, whereas the lateral condyle size slightly reduces in posterior direction [23]. In addition, there is a groove for the popliteus insertion in the lateral femoral epicondyle [24], which can be utilized straightforwardly for identification.

The tibial plateau, which has asymmetrical oval, concave medial and circular, convex lateral compartment, is divided by the intercondylar eminence (IM) [17, 18, 22]. In front of IM, there is tibial tuberosity, which is a low elevation of tibial outline for the insertion of quadriceps tendon [25]. Tibial plateau is covered by articular cartilage. The lateral plateau is smaller than the medial plateau [22], corresponding to the fact that the dimension of the medial femoral condyle is larger than the lateral. Since it is well recognized that the anatomical feature of the joint is to satisfy the fundamental physiological function, different geometrical characteristics at the medial and lateral sides of both the femur and the tibia facilitate the multiple relative motion of the knee joint, directly determining knee kinematics and kinetics.
The patellofemoral joint is a sellar joint between the patella and the femoral trochlea (sulcus), and its major role is to increase the mechanical advantage of the extensor mechanism. The normal angle of the femoral sulcus is 137°, with a variation of 8° [26]. Furthermore, the notch on the lateral femoral condyle separates the patellofemoral area from the tibiofemoral joint contact area, while for the medial condyle, similar mark is not available. Geometrical congruence can be found between the femoral trochlear groove and the articular geometry of the patella [23]. The femoral trochlear groove is flatter at the proximal portion than more distally, which could bring about more risk of patellar lateral subluxation at the stage close to full extension than during flexion, considering that the sulcus deepens distally and provides greater geometrical conformity and constraints for patellar tracking [19]. The trochlear groove also has a broader, steeper surface and extends more anteriorly on the lateral side than medial, which reveals that better patellofemoral articulation is on the lateral side.

The articular surface of the patella takes the proximal two thirds of the total bony area. There are seven patellar facets within the articular surface of which 3 located medially (superior, middle and inferior facets), 3 laterally and one extra, non-articulating facet on the medial side (the odd facet). The medial facets are slightly smaller than the lateral in overall dimensions, which implies that the lateral facets are the predominant portion compared with medial in determining tracking kinematics. The medial facet surfaces are also more convex in shape [19, 27].

Based on the particular geometry, along with the knee flexion, the initial patellofemoral contact forms laterally and the direction of the patellar tracking is from
proximomedial to distolateral, while in the maximally flexed knee, the traction is from proximolateral to distomedial [23]. During motion, the patellar stability is maintained through the combination of bony, ligamentous, and muscular restraints [18].

1.3.3 Muscles and Ligaments

Within this section, anatomy and geometry of muscles and ligaments were briefly discussed.

Regarding the medial compartmental structure of the knee, it can be divided into 3 discrete layers [18, 25, 28, 29, 30]. The most superficial layer is the deep or crural fascia, which is the first plane underneath the subcutaneous tissue, and extends from the patella to the midline of the popliteal fossa [18, 29]. Anteriorly, this layer blends with the Layer II in a vertical line, 1 to 2 cm anterior to the anterior edge of the superficial medial collateral ligament (MCL). Posteriorly, Layer I overlies the 2 heads of the gastrocnemius, superficial to the tendons of the semimembranosus, semitendinosus and gracilis muscles, and serves to support the neurovascular structures of the popliteal fossa [30, 31].

Layer II (intermediate layer) contains principally the superficial medial collateral ligament. Superficial MCL and deep MCL (Layer III) are the majorly two constitutional components of medial collateral ligament, separating anteriorly from each other by an intermediate signal intensity interface (probably fat and intervening bursa), while merging with each other posteriorly [25, 29, 32, 33]. The superficial MCL composes of vertically-oriented and obliquely-oriented fibers [30, 33]. Posteriorly, the oblique portion of the MCL is found to fuse with Layer III and closely attached to the posteromedial part of the meniscus [29, 30, 33, 34]. Along the posterior aspect of the knee, this combined
structure merges fibers from the semimembranosus tendon and the synovial tendon sheath, forming the oblique popliteal ligament [18, 30, 33]. In addition, from the region of the femoral insertion of the anterior fibers, a transverse band runs forward in the plane of Layer II from the adductor tubercle toward the superomedial aspect of patella, forming the medial patellofemoral ligament, which is a distinct condensation of capsular fibers in coronal plane [18, 25, 29]. As mentioned before, at the anterior margin of the superficial MCL, Layer II is split in a vertical plane with fibers which run anteriorly joining Layer I [33], and merging with medial patellar retinaculum (which is extension of vastus medialis muscle passing the patella, while that of vastus lateralis makes lateral retinaculum) and patellofemoral ligament [25].

Layer III (deep layer) refers to the deepest capsular layer, which is firmly and uniformly attached to the medial meniscus [18, 25, 29-34]. This layer is very thin and redundant anteriorly to accommodate the range of motion of the knee [18, 25], yet beneath the superficial medial collateral ligament the fibers of Layer III thicken, forming the deep portion of MCL locating close to the meniscus, which is also the middle capsular [22, 30, 33]. The deep portion of the MCL is composed of the meniscofemoral and meniscotibial extensions [30]. The meniscofemoral portion of the deep ligament extends from the femur to the mid-portion of the peripheral margin of the meniscus. The meniscotibial portion of the ligament anchors the meniscus, and is separated from the overlying superficial MCL. The deep MCL is a major secondary restraint to anterior translation [25].
Regarding the lateral compartmental structure of the knee, it is reported that similar layer-based structures also exist. Analogous to the medial aspect, these structures include the superficial (I), intermediate (II), and capsular layers (III) [33, 35].

Layer I is formed by the iliotibial tract (IT) with its anterior extension, and the biceps tendon with its posterior expansion. Layer I extends from the prepatellar bursa (at anterior margin of the patella) to the posterior aspect of the popliteal fossa, which follows the same pattern as the medial compartment. Layer I blends with layer II anteriorly to form a more distinct and thicker band [18, 25, 33, 35].

One of constituents in Layer II is the lateral collateral ligament, which originates from the femoral epicondyle to the fibular head, and is overlain by the superficial lamina of Layer III. Anteriorly, layer II is formed by the retinaculum of the quadriceps, and due to its incomplete structure, it is represented posteriorly by the lateral patellofemoral ligaments [25, 33, 35]. Also, as part of layer II, the patellomeniscal ligament travels obliquely from the patella to the margin of the lateral meniscus, and ends at Gerdy's tubercle [18, 25, 35]. Layers II and III are adherent to each other in a vertical line at lateral margin of the patella [25].

Layer III, analogous to the medial compartment, is the deepest layer and attached to the lateral edges of the tibia and femur circumferentially in horizontal planes. At the proximal and distal ends, it composes two laminae posterior to the IT band. The more superficial lamina encompasses the lateral collateral ligament (LCL) and ends posteriorly at the fabellofibular ligament [18, 33, 35]. And the deeper lamina passes posteriorly along the lateral edge of meniscus, forming meniscotibial ligament, which is the capsular
attachment to outer meniscal edge, and hiatus for passage of popliteus tendon [18, 25, 35]. The bifurcation of layer III into two laminae is identified posteriorly behind the overlying iliotibial tract. Anteriorly, Layers II and III are separated clearly from each other by intervening adipose tissue [18, 25].

Collateral Ligaments

The MCL is composed of a superficial and a deep portion. The superficial MCL originates on the medial femoral epicondyle, and runs downward as a broad triangular band approximately 11 cm to its tibial insertion, deep to the gracilis and semitendinosus tendons [18]. The superficial MCL can be further subdivided into anterior and posterior portions. The anterior portion extends vertically, whereas the posterior margin passes obliquely backwards to an insertion in the medial meniscus. The deep portion of the MCL also can be divided into 2 subdivisions, the meniscofemoral and meniscotibial ligaments, defined by their respective insertions [28, 29, 33, 34].

The LCL is a round, sharply defined, pencil-thick band which extends from the lateral femoral epicondyle to the head of the fibula [23]. Average length of the ligament is reported from 59.2 to 71 mm [36] and has a minimum diameter at its midpoint, where it is elliptical in shape. The LCL arises in a fan-like fashion in a fovea immediately posterior to the lateral epicondyle at an average 3.7 mm posterior to the apex of the epicondylar tubercle [18]. Also, it is superficial to the tendon of the popliteus [36, 37]. The fibular attachment of LCL is into a superior and laterally facing V-shaped plateau on the head of the fibula [36]. Compared with the MCL, the LCL is distinguished from based on the fact that it is separated from the lateral meniscus [22, 23, 25].
Medial Patellofemoral Ligaments

The medial patellofemoral ligament is recognized as a major restraint to lateral displacement of the distal knee-extensor mechanism, which plays a critical role in knee joint function [18]. As described above, the MPFL, which is hourglass-shaped, runs transversely in Layer II from attachments to the adductor tubercle, as well as femoral epicondyle, and anterior border of the superficial MCL [29-33]. The proximal fibers of the ligament proceed anteriorly toward to the vastus medialis (obliquus), diverging proximally to insert to the undersurface of the vastus medialis obliquus and aponeurotic fibers of the vastus intermedius. The distal fibers insert anteriorly to the superomedial patella, extending inferiorly from the medial aspect [31, 38, 39]. The width of the medial patellofemoral ligament is in average 1.3 cm.

Iliotibial Band

The IT band is formed proximally at the level of the greater trochanter by the coalescence of fascial investments of the tensor fascia lata, gluteus maximus, and gluteus medius. It then attaches to the femur through the lateral intermuscular septum. At the knee, it separates into 2 functional components: the iliotibial tract, and iliopatellar band. The iliotibial tract originates from supracondylar tubercle of the femur, which has 3 layers: the capsuloosseous, deep, and superficial layer, and merge distally to the lateral tibial tuberosity, posteriorly and proximally to Gerdy's tubercle [24, 37, 40]. The iliopatellar component of the IT tract connects the anterior aspect of the IT tract to the patella, which functions to resist a medially-directed force exerted on the patella, and is dynamically affected by the vastus lateralis [18, 40].
Extensor Mechanism

The extensor mechanism consists of the quadriceps muscle group, the patella, quadriceps tendon, and tibial tubercle. This mechanism is functionally responsible for extension of the knee. The quadriceps muscle group (6 muscles totally, which can also be summarized to 4 muscles: the rectus femoris, the vastus lateralis, the vastus medialis and the vastus intermedius) joins in a trilaminar fashion to form the quadriceps tendon [19]. The rectus femoris originates from two attachments on the ilium, which runs in the anterior thigh underneath the sartorius and joins the quadriceps tendon 3 to 5 cm above the patella. The vastus lateralis originates from the lateral aspect of the femur and merges into the quadriceps tendon through the lateral retinaculum 3 cm from the superolateral aspect of the patella. The vastus medialis arises from the superomedial aspect of the femur and includes two groups of fibers, the medialis obliquus and medialis longus. The vastus intermedius lies under all three muscles, with tendinous fibers blending medially and laterally with the vastus medialis and vastus lateralis, and inserting distally into the superior edge of the patella [19, 22-24]. The fibers of the quadriceps tendon extend across the anterior surface of the patella and blend distally with the quadriceps tendon.

The patellofemoral ligaments provide stabilized force for patellar tracking along the femoral sulcus/trochlear groove. Also, the lateral patellofemoral ligament, which is a similar functional representation of the lateral retinacular structure, is located superiorly and inferiorly on the lateral knee compartment, extending from the anterior surface of the patella to posterior side of the lateral femoral condyle [19].
The distal portion of the quadriceps tendon extends from the inferior pole of the patella to the tibial tubercle, with dimension of 25 to 40 mm wide and 4 to 6 cm long in average adults [17, 19]. It inserts into the proximal pole of the patella and directs the pull of the quadriceps muscles. The orientation of the quadriceps tendon, as well as the resultant quadriceps muscle force, is defined based on the Q-angle. When knee is fully extended, the Q-angle is determined as angle between the connecting line runs from the center of the patella and the quadriceps tendon attachment site to the tibial tubercle, and the connecting line between the center of the patella and the anterior superior iliac spine on the pelvis [41].Normally, this angle varies between 6° and 27°, with a mean value of approximately 15° [42]. It has been well noticed that the magnitude of Q-angle influences the biomechanical properties of the patellofemoral joint. Based on previous research studies, both in-vitro experimental testing [43] and theoretical modeling evaluation [44] have shown that increasing the Q-angle tends to increase the lateral patellofemoral contact pressure, while decreasing the angle tends to increase the medial patellofemoral contact pressure.

**1.4 Knee Joint Kinematics and Kinetics**

**1.4.1 Tibiofemoral Joint Mechanics**

Regarding the overall knee joint mechanics, for clear description, all loads are expressed within the 3D spatial coordinate system. In detail, the tibial axis (proximal-distal) and two axes perpendicular to it in the anterior-posterior and medial-lateral directions are often recruited to construct the reference frame.
Theoretically, based on the 3D physiological coordinate system, a force in the direction of the proximal-distal direction resists inter-penetration of the bones when in compression, distraction of the bones when in tension. Forces exist in the anterior-posterior and medial-lateral directions to resist (or drive) relative translations of bones. A moment about the medial-lateral axis resists (or drives) flexion or extension. A moment about the anterior-posterior axis resists (or drives) abduction or adduction (varus and valgus). And lastly, a moment about the proximal-distal direction resists (or drives) medial and lateral rotation. Due to the smooth articular surface which provides minimum level of resistance to sliding movements from one bone to the other, shear stress could not be introduced with the translational movement. Also, the unavailability of adhesion determines that inter-surface tension will not be involved between articular bones. In general, the aforementioned loading components are applied and transmitted majorly from the axial compression between the articulating bones at the ends of the shafts, and the combination of tension by fibers of the ligaments and tendons within the knee joint [45].

The local kinematic components are initiated and limited/controlled by the lower limb biomechanics. Considering the most fundamental daily activities, variation of loads exerted on the knee is influenced by the position of center of body weight (BW), ligament restraining and muscle contraction, as well as joint/ground reactions, etc. To simplify the analysis, asymmetrical stance, or one leg standing, which is the elemental action of the dynamic gait process, is taken into consideration compared with the symmetric stance.
In the standing position on both feet, both knees support the part of the body above, and the BW can be simply treated as a concentrated force vector at the center of gravity, locating at the level of the third lumbar vertebra. Projected on the coronal plane, the center of gravity and joints are at the same straight line passing through the ground reaction force. Also, on the sagittal plane, the center of gravity also lies close to the vertical line passing through the flexion centers of knees. For such mechanical system, the BW is evenly distributed between both knees at the vertical direction, and knee muscular force/ligamentous force could be negligible since the force equilibrium has already been accomplished.

In contrast, in standing position on one foot, the loaded knee supports the head, the trunk, the upper limbs, the loaded thigh and the opposite lower limb. The partial body weight (93% of the total BW) is concentrated at a different point position compared with the previous case. On the coronal plane, in contrast, action line of the partial BW (denoted by \( P \)) is not centrally-aligned but medially-inclined, which introduces a lateral force \( L \), which is muscular tension generated to balance the moment caused by body gravity to prevent tilting of the femur. In detail, force \( L \) is constituted by the gluteus maximus, the tensor fasciae latae and the iliotibial band. Regarding the knee reaction, with fulfilling of the moment equilibrium, the resultant force \( R \) combing \( P \) and \( L \), is exerted between the curvatures of medial and lateral femoral condyles, with approximate 5º inclination about the vertical direction. \( R \) can only be neutralized by the knee joint reaction force, ultimately the ground reaction.
Within the sagittal plane, for one leg stance posture with the center of body weight that translates more anteriorly than the original, a more complicated interplay of forces occurs when knee flexes [46]. The partial body weight $P$ is applied along a vertical line through the forefoot, which is the support of the entire lower limb. $P$ creates dorsiflexion of the foot, which can be counterbalanced by the activation of calf muscle, thus generating tension $M_c$ for moment equilibrium. Yet $P$ and $M_c$ lead to corresponding ground reactions, which form the resultant force $R_1$ lining through the flexion center of the ankle for maintaining force equilibrium. Furthermore, $P$ tends to rotate the pelvis anteriorly, whereas hamstring muscles provide combined force $M_h$ to compensate the tilting moment. $P$ and $M_h$ are combined from the standpoint of force balance, which results in $R_2$, passing through the center of the femoral head but posterior to the knee. For the knee joint, $R_2$ increases the rotation towards to the counterclockwise direction, accordingly, $M_g$ is produced by gastrocnemius to counterbalance $R_2$ at the back side of the knee, by which further resultant force $R_3$ is also developed behind the joint rotating axis, remaining flexing the knee. A force in front of the knee is thus required for moment equilibrium. Quadriceps tendon (distal portion) provides tension $P_p$, and by mathematically combined with $R_3$, both lead to the resultant force $R_4$ to preserve the tibiofemoral joint stability. However, as it has been mentioned, the patellofemoral joint cannot be ignored at this site. Since the distal portion of the quadriceps tendon is tighten and activated, with respect to the patella, which serves as a fulcrum, the PF joint will be hardly balanced unless the proximal portion of the quadriceps tendon is also tensed, presenting force $M_q$. $P_p$ and $M_q$ satisfy the following equation:
\[ M_q \cdot q = P_p \cdot k \]

(1.1)

where \( q \) and \( k \) are moment arms of \( M_q \) and \( P_p \) relative to the knee flexion center, respectively.

In addition, \( M_q \) and \( P_p \) form the resultant force \( R_5 \), which is the source of the compressive force in patellofemoral joint, while \( R_4 \) creates the tibiofemoral joint compression [46]. Within the sagittal plane, the whole multi-body system follows the force/moment equilibrium and the static kinematic conditions, so that the muscular/ligamentous contraction can be systematically determined [47]. In the analysis explained above, muscular/ligamentous forces are simplified and reduced numerically to avoid redundancy in mathematical calculation. The redundant muscles/ligaments are inevitable in passive contraction, thus the central nerve system organizes this biological process based upon advanced optimization algorithm to lower average level of muscle activity within contraction envelopes.

As discussed above, the knee joint mechanics fundamentally refers to rigid body dynamics, combing with ligamentous and muscular force interaction. It has been summarized that the mechanisms for maintaining the joint stability and force/moment equilibrium are passive joint reaction, active muscle force (musculotendinous units) and passive ligament force [45, 48, 49], where “passive” means constraints imposed due to stiffness of the ligaments, and the geometrical congruence of the joint surfaces, while “active”, on the contrary, majorly refers to muscle contraction and shortening [45, 48].

Regarding the tibiofemoral joint kinematics, flexion/extension is the primary motion type. Flexion motion is majorly induced by the flexor muscle contraction and
resisted actively by the couple consisting of tension in the distal portion of the quadriceps tendon and compression between two articular surfaces against interpenetration (force and orientation are relative to the femur), with the distance between the action line of the quadriceps tendon at distal insertion site and center of the pressure within the contact area as the lever arm [45, 48]. The active resistance force triggers passive joint reaction force for joining femur and tibia functionally. Due to geometrical inconformity, curvatures of the articular contact surfaces provide insufficient compressive force throughout range of motion to maintain joint force equilibrium. Moreover, frictional resistance is limited on synovial surfaces. Therefore, both factors lead to an unstable joint, and a limited range of anterior-posterior gliding of articular surfaces occurs in all flexed/extended positions except full extension. Relative femoral sliding anteriorly on the tibia plateau during flexion stretches the posterior cruciate ligament and thereby creates gradually increasing passive tension within it, which presents the missing component of the joint reaction force [48-50]. Ultimately, balance is reached so that further translational movement can be ceased when the component of the soft tissue tension which is parallel to the articular surfaces equals to the applied shear force that is introduced by flexion. Hence, the range of anterior-posterior sliding motion depends on the extent of ligaments deformed in response to the rigid body motion of bones, as well as the relation between the geometry of the articular surfaces and disposition of the soft tissues [45]. Due to not only the sliding movement but also the anatomical feature of the cruciate ligaments and the collateral ligaments, an obligatory rolling movement predominates in the initial 30° of flexion which carries the tibiofemoral contacting points backwards on the tibia.
(reference) in flexion and forwards in extension based on the pattern of the cam-shaped femoral condylar structures [18, 22, 23, 45]. It was found that the backward/posterior movement of the contact area in flexion tends to increase the power of the quadriceps to extend the knee, and simultaneously decrease the compressive force at the articular surfaces by increasing the lever arm of the couple mentioned before [45, 48].

Muscles associated with the joint actively function as primary movers. Specially, McLeod et al stated that the two synergistic muscle groups of the musculotendinous: the quadriceps femoris group and the hamstrings, have distinct or opposite physiological focuses. The former is responsible for extension of the knee and deceleration of the forward motion of the femur on the tibia, and the latter stimulates flexion of the knee and internal-external rotational motion of the femur on the tibia [49].

In all positions except full extension during knee flexion/extension, a range of internal-external rotation is available, which ranges from about 30° internally to a few degrees externally. In addition, fully extended joint will not rotate since the tightened soft tissue structure spanning across the joint fully restrains the knee to prevent rotation.

Furthermore, the tibial eminence provides secondary mechanism in constraining and controlling rotation. Location of the medial/lateral joint contacting area on the tibial plateau moves alongside with the femoral internal-external rotation, since one side displaces posteriorly while the other anteriorly [45]. Compressive tibiofemoral forces on both sides of the tibial plateau could provide planar force couple to balance the applied torque from hamstring activation, even the effect is limited. The lever arm of the couple
arises from the relatively anterior position of one contact area and the posterior position of the other [45].

Another type of the knee joint motion, abstraction and adduction, are induced by medial-lateral muscular force exerted by abductor and adductor muscle groups, which translate tibia and femur relative to each other in the medial-lateral direction. To balance this force, an equal medial-lateral force at the opposite direction and a moment about the anterior-posterior axis are transmitted across the joint [45]. Such balance/resistance force mainly originates from the collateral ligaments, in together with compression on the lateral condyle. Besides, the cruciate ligaments can also provide passive rotational constraint, which has a shorter moment arm than the collateral ligaments.

When the adduction motion occurs about the rotating center which locates within the contact area of the medial femoral condyle, the tensile forces at the lateral compartment of the knee contribute to moment resisting the varus rotation with respect to the fulcrum. In detail, tension in collateral ligaments and cruciate ligaments is recruited to compensate the adduction movement. To balance the applied moment, which is denoted by F*a about the center of tibiofemoral joint, the tensile force provided by ligaments must be several times larger than the applied medially-inclined force in magnitude due to shorter moment arm [45, 48, 49]. Also, from the perspective of force equilibrium, leaning of the collateral ligament (about 10°) due the applied load from adductor muscles could generate horizontal component of the tensile force to preserve the equilibrium along the horizontal direction. Adduction motion also implies medial-lateral translation of the
bones and the resultant contact force could be transmitted partially by the tibial eminence, assisting in poising the applied load [45, 48, 49].

To measure the accurate level of the TF joint reaction to appropriately direct patients’ rehabilitation, previous researchers have utilized various techniques to measure the tibiofemoral joint reaction force after TKR. According to the observation conducted by D’Lima et al, by the 6th postoperative week, the peak tibiofemoral axial force during walking was in average 2.2 times of body weight (BW). And during the stair climbing, the loading increased from 1.9 times BW on Day 6 to 2.5 times BW at 6 weeks [51]. From the modeling viewpoint, such experimental measurements would be beneficial in determining the accurate boundary conditions and calibrating predicted results.

**Biomechanical Function of the Ligamentous Structures**

According to the research conclusion drawn by Wilson et al., ligaments, and the articular surfaces, principally guide the passive knee flexion, indicating that ligamentous structures maintain to provide required mechanical constraints and support in knee motion [52]. Since the tibiofemoral articular surfaces are not fully conformal, ligamentous structures serve as major components for passive resistance to retain the force and moment equilibrium. For more details, several major ligamentous structures are further introduced below.

**Cruciate ligaments:** the anterior cruciate ligament is a essential to stable and normal knee joint flexion and rotation. It is the primary restraint to anterior translation of the tibia relative to the femur (about 80% contribution), and a secondary restraint to internal femoral rotation, varus and valgus tibiofemoral rotation, as well as
hyperextension [53-56], ACL deficiency may result in disintegration of the gliding movement of the femur relative to tibia. This ligament also plays a vital role in controlling internal tibial rotation which combines anterior translation during the anterior-posterior movement [57].

The posterior cruciate ligament is the primary and only ligamentous restraint to posterior translation of the tibia relative to femur during flexion (90% contribution), and a secondary restraint to tibiofemoral varus and valgus rotation, as well as femoral external rotation [53-57]. Though the external rotation is not directly restrained by cruciate ligaments, PCL could be expected to provide restraint when it becomes taut at 90° of flexion [25]. Sectioning of ACL or PCL will in general alter the pattern of the femoral and tibial rotation and relative translation, causing instability and unexpected ligaments dysfunction.

Collateral ligaments: The medial collateral ligament is a crucial constraint to abduction, internal rotation and medial-lateral translation of the tibia. As mentioned in the previous sections, the MCL constitutes superficial and deep layers, of which the former is a major contributor to joint stability, which provides large portion of mechanical control for internal femoral rotation [25], and also the key restraint to valgus tibial rotation, with contribution estimated from 57% at 5° to 78% by 25° knee flexion [58, 59]. The functional significance of the MCL increases along with increased flexion, as the posterior capsular structural constraint becomes slack at deeper flexion, indicating clinically that excessive laxity and joint space widening could be expected following MCL deficiency, especially at higher flexion angles [18].
The lateral collateral ligament, located posteriorly to the femoral flexion axis, is the primary restraint to tibial adduction during knee flexion and a secondary restraint to external femoral rotation and posterior translation. It is tightest at full extension and progressively relaxes at flexion angles beyond 30° [36, 59-62]. To resist varus instability, LCL remains taut from 0° to 30° flexion for sufficient tensile force, yet external femoral rotation could be allowed and the constraining effect on varus motion decreases when LCL slackens beyond the range [36]. In addition, the anterior-posterior translation and internal rotation are generally free of LCL control.

1.4.2 Patellofemoral Joint Mechanics and Patella Kinematics

Simultaneous motions occur during the knee flexion, including the TF and PF joint movements. Changes of relative position and orientation of femur and patella result in variation of patellar kinematics and function. As experimental observations reveal, mechanical contribution of the patella, which depends on the elevation of quadriceps tendon due to patella, is found to be much smaller in deep flexion than in full extension.

In deep flexion, the patella sinks into the intercondylar notch and produces little anterior displacement of the quadriceps tendon. Therefore, little effect on the extensor moment arm can be expected. Towards to the full extension, patella rises out of the intercondylar groove and produces significant anterior displacement of the quadriceps tendon, resulting in the rapidly increased extensor moment arm [19].

Within the patellofemoral joint, based on tension in proximal and distal parts of the quadriceps tendon, \( F_q \) and \( F_p \), the patellofemoral joint reaction (PFJR) force, which is perpendicular to the point of PF joint contact, can be calculated as:
\[
PFJR = \sqrt{F_q^2 + F_p^2 + 2F_qF_p \cos \gamma}
\]  \hspace{1cm} (1.2)

where \( \gamma \) is the angle between the distal and proximal quadriceps tendon force vectors.

The PFJR force is the resultant force of tension in the distal and proximal portions of the quadriceps tendon. Practically, rehabilitation programs that are designed to minimize the PFJR force should avoid a range of motion between 70º and 120º, during which the PFJR force is 100\% of the quadriceps force [63]. Also, based on the triangular relation, PFJR force can also be calculated as:

\[
PFJR = 2F_q \cos \left(\frac{\gamma}{2}\right)
\]  \hspace{1cm} (1.3)

In detail, within the knee, the tibiofemoral rotation axis does not coincide with the center of patellofemoral joint, which results in two separated sets of moment equilibrium about both centers [46]. During the squatting posture, the resultant PFJR vector increases with the increased flexion angle, attributing to closing of the angle formed by the action lines of tensions in the proximal and distal parts of the quadriceps tendon [19, 46].

Previous studies have provided several references about the PFJR force magnitude. Reilly et al studied the PFJR force during level walking, stair climbing, straight leg raising exercises and deep knee bends. Force magnitudes were recorded and calculated: 0.5 BW for level walking, 7.6 BW for deep knee bends, 3.3 BW for stair climbing or descending, 1.4 BW at 36º flexion during the knee extension exercise, and 0.5 BW for straight leg raising exercise [64]. For different activities, the patellofemoral joint reaction may vary greatly based on different external loading conditions, muscular and ligamentous controls.
The patellofemoral joint reaction force, in the form of joint compression, directly induces the patellofemoral joint contact. Contact pressure and contact area are determined by the interrelationship between the patellar and femoral geometrical/kinematic characteristics. Specifically, patella always enters the trochlea from the lateral side because of the Q-angle. Once the patella-trochlea contact is made, the resultant flexion compresses the patella against the femur. And the center of the contact area travels from the patellar distal to the proximal end, thus the patella contacts with femur at its distal end at small flexion \([65]\). Continuously, by 30\(^\circ\) of flexion, the contact area is evenly distributed on both sides of femoral condyles. At 60\(^\circ\) of flexion, effective contact is formed between the central portion of the patellar surface and the femoral groove. At 90 of flexion, the contact is between the superior aspect of the patellar articular cartilage and an area close to the proximal end of the femoral groove just above the notch. At 120\(^\circ\) flexion, contact occurs between the superior aspect of the patella and two areas surrounding the Intracondylar notch of the femur \([42]\).

During the first 30\(^\circ\) of knee flexion, effective patella-femur bone contact is not formed, which allows multiple degrees of freedom. This stage leads to the least stable patella \([27]\). To maintain the overall patellar stability, the slope of the lateral facet of the trochlea provides the most considerable mechanical limits and balance force once the patella engages into the trochlear groove through the congruous geometrical constraints, whereas the retinacular structure and ligamentous restraints are also important for patellar stability. Hence, articular geometry, muscle action and passive ligamentous/soft-tissue control are the three major factors affecting the overall patellar stability. The patella
responds to a set of 3 forces: the pull of the quadriceps, hamstrings, and a net compressive force on the patellofemoral surfaces. In addition, soft tissue constraints contribute to the tracking of the patella within the trochlear groove include the medial patellofemoral ligament, medial patellomeniscal ligament, medial patellotibial ligament, medial retinaculum, and lateral retinaculum [18].

Patellar motion driven by knee flexion includes 6 degrees of freedom: medial-lateral translation, superior-inferior translation, anterior-posterior translation, medial-lateral rotation, medial-lateral tilt, and flexion-extension. The alignment of the femoral trochlear groove and the patella follows the dictated orientation of sulcus axis, which deviates laterally from the anatomic axis of the femur within the coronal plane [66]. To assist stabilization, the quadriceps muscle, quadriceps tendon, and medial and lateral retinacular structures mechanically keep the patella in position. As flexion increases, the geometrical reaction provided by the trochlear groove takes the major portion of kinematic constraints.

During the tracking motion, according to the observation by Hsu et al, patellar flexion angle in the intact knee increases along with flexion angle increase [65], yet it tends to lag behind tibiofemoral knee flexion [27]. Also, the patella shows a progressively increased lateral, posterior, and distal shift with a lateral tilt and internal rotation as the knee flexion angle increases [65]. Many studies also described initial medial tilting before the lateral tilt progressively accumulates [67]. Clinically, it is significant to well preserve normal tendon length (avoid laxity and dysfunction) and alignment during the TKR surgery [27, 68].
CHAPTER 2 RESEARCH OVERVIEW

2.1 Overview

This PhD dissertation focused on the field of tibiofemoral and patellofemoral biomechanics following total knee replacement. The ultimate goal of this work was to elucidate the intraoperative effect of the patellar component thickness on parameters associated with the knee joint biomechanics, particularly the patellar kinematics, quadriceps tendon force and patellofemoral joint contact pressure. To select appropriate research solution and generate comprehensive understanding about patellar biomechanics, within the current project, three steps were taken to progressively study the problem.

Firstly, the biomechanical interaction between the tibiofemoral and patellofemoral joint was identified. Based on clinical practice, patellar tracking and patellofemoral biomechanics are greatly affected by tibiofemoral alignment during TKR surgery. Therefore, for the first step, combined use of experimental knee simulator testing and computational finite element (FE) modeling was recruited to explore the effect of TKR rotational alignment variation, and identify the potential impact on patellofemoral joint and extensor mechanism.

Secondly, preliminary attempt was made to investigate the relationship between variation of the patellar component thickness and patellofemoral biomechanical parameters, including the quadriceps tendon force and patellofemoral contact pressure. A 3D human knee joint FE model was established with realistic geometrical/anatomic
representation, material property characterization and dynamic displacement boundary conditions acquired from experimental simulator testing. The patellar component was geometrically morphed to various thickness levels to generate a series of FE models for comparative analysis.

Thirdly, based on previous research effort, a patient-specific cadaver study was further conducted to comprehensively explore the biomechanical influence of patellar component thickness on detailed patellar kinematics, quadriceps tendon force and patellofemoral contact pressure. This testing utilized different instruments for real-time patellar motion capture and patellofemoral contact pressure measurement. As a result, this study could provide an advanced reference tool for clinical practitioners in determining appropriate patellar component during TKR surgery.

2.2 Specific Aims

Specific aims of this study include:

a. Complete a patient-specific, 3D TKR knee joint model which consists of major biological components.

b. Complete the computational description of the material behaviors of biological tissues and TKR implant components

c. Conduct a quasi-static finite element analysis for investigation of effect of patellar component thickness for $0^\circ$-$60^\circ$ knee motion

d. Conduct a cadaver experiment to assess the effect of patellar component thickness on patellar kinematics, quadriceps tendon force and patellofemoral contact pressure
Figure 2.1 Schematic flow chart of the current project
CHAPTER 3  EFFECT OF ROTATIONAL PROSTHETIC ALIGNMENT VARIATION ON THE TIBIOFEMORAL CONTACT PRESSURE AND JOINT KINEMATICS IN TOTAL KNEE REPLACEMENT

3.1 Introduction

During total knee replacement (TKR) surgeries, prosthetic alignment is considered as one of most important issues determining the implant longevity and long-term survival rate [69, 70]. Postoperative femoral-tibial axial alignment determines resultant joint kinematics, stress distribution along the contacting surface and surrounding soft tissue deformation [71-76]. Inappropriate tibiofemoral alignment could cause undesirable clinical outcomes, especially postoperative complications and revision TKR surgeries [77, 78]. Specifically, tibiofemoral alignment is essential to maintain appropriate patellar tracking and joint space during knee flexion/extension, normally-tensed soft tissue structures and joint stability [79]. Knee alignment is generally determined at both the coronal and sagittal plane, with several variables specified during TKR surgery, including the tibiofemoral varus-valgus (V/V) rotational alignment angle, femoral internal-external (I/E) rotational alignment angle, tibial posterior slope, etc.

Physical estimation noticed that greater applied knee joint loading is transferred through the medial knee compartment than the lateral compartment during daily activities [80], which contributes to much more medial compartment osteoarthritis (OA) incidences.
seen in the clinical practice [81-83]. Compared with the TKR surgical technique for medial compartment OA, additional dynamic ligament rebalance tends to be crucial to maintain normal patellar tracking and pain release for lateral compartment OA cases.

The commonly-accepted surgical protocol for most TKR cases is defined following the lower limb mechanical axis, which draws from the center of the femoral head to the center of the ankle joint [84], and makes an angle of 5-7° with the anatomical axis along the femoral and tibial intramedullary direction [85, 86]. Thus, distal femoral cut with an angle of approximately 5° relative to the anatomical axis is preferred in the current TKR alignment procedure [87], to maintain the postoperative tibial neutral orientation and avoid excessive varus or valgus tibiofemoral orientation. For I/E alignment of the femoral component, the Whiteside’s line, which runs along the deepest curvature of the trochlear groove [88] and perpendicular to the transepicondylar axis, is widely deemed as the most reliable reference, though its repeatability and precision as a rotational alignment reference has been questioned by previous investigations [89, 90].

Particularly, for most medial compartment OA cases with narrower joint space at the medial side, 3° external femoral rotation is introduced via posterior femoral cut, to further release the over-tensed medial soft tissue structures due to the preoperative varus tibiofemoral orientation [87]. By inspecting soft tissue release and femoral I/E rotation, the tibial component is accordingly adjusted and generally aligned based on the medial third of the tibial tuberosity as a reference adopted in most clinical practices [85, 87].

Depending on TKR insertion protocols specified by different implant manufacturers,
tibial posterior slope is commonly set to 3-7° [91, 92], to extensively facilitate the deep knee flexion but maintain adequate motion stability [93].

Prevalent surgical protocol for TKR alignment has been constantly improved along with progress of medical technology. However, several variables in prosthetic alignment still remain questionable, e.g. effective I/E rotational reference for tibia and femur with lateral compartment OA is lacking; also, due to complexity of patients’ specific physical conditions, accurate control for alignment is hardly achieved. Thus postoperative TKR alignment variation has been constantly identified in previous studies. Mahaluxmivala et al. noted that in 673 TKR cases, standard deviations of 2.3° and 3.5° existed for V/V angle and tibial slope, respectively [92], and Zihlmann et al. further summarized that 6° internal to 8° external variance for femoral I/E alignment occurred in 10-30% of TKR patients [94]. Furthermore, previous effort regarding the influence of alignment variation on postoperative knee function and clinical outcome has been made via in-vitro or in-vivo studies. Ritter et al. and Fang et al. both concluded that postoperative knees with <2.5° and >7.4° valgus angle had significantly higher risk of failure [95, 96], and thus conventional clinical thoughts recognized that the V/V deviation range for the tibiofemoral mechanical axis within the coronal plane is ±3° for desired postoperative outcome [97-99]. It has also been found that excessive valgus tibiofemoral alignment in the coronal plane tends to increase the laxity of the lateral knee compartment, generating an inclined joint line [79]. Additionally, excessive femoral internal rotation in alignment possibly leads to lateral condylar lift-off at 90° of flexion. Inappropriate external rotation is associated with medial condylar lift-off during knee
flexion [79], and more severely, patellar maltracking and instability [100] due to abnormal medial-lateral translation of the patellar tendon insertion site during knee joint motion. Moreover, tibiofemoral joint stability within the sagittal plane is closely relevant to the tibial slope, which could cause unexpected alteration of the joint space [79]. Specifically, increased tibial slope could contribute significantly to anterior-posterior deflection [101], while on contrary, the correlation between tibial slope and PCL overextension has been demonstrated by Singerman and collaborators through in vitro testing [102], indicating the importance of tibial slope for TKR alignment especially when cruciate retaining (CR) design is implanted.

From an engineering perspective, the altered clinical outcome can be more precisely depicted and reflected in terms of equivalent quantifiable measures such as joint loading distribution, tibiofemoral contact pressure and resultant relative knee kinematics, which are ultimately associated with the long-term wear rate and fatigue life of the implant [70, 103, 104]. As a widely-used testing device for assessing TKR kinematic and wear performance, the force-controlled Instron-Stanmore knee simulator (Instron Corp., Canton, MA) is capable of quantitatively predicting postoperative TKR kinematics and wear propagation under well-controlled experimental conditions, which is more efficient and flexible than regular in-vivo or in-vitro testing. Using the Instron-Stanmore simulator, Laz et al. and Easley et al. established a probabilistic evaluation platform using computational finite element analysis to identify individual sensitivity of the TKR alignment variables [105, 106] in terms of dynamic tibiofemoral contact pressure. However, parametric analysis with variation range from realistic surgical practices is still
lacking of major rotational alignment variables. Therefore, the current study utilized validated FEA models with the same experimental settings and boundary conditions as the Instron-Stanmore knee simulator to calculate postoperative TKR motion and real-time tibiofemoral contact pressure distribution during a walking cycle. Alignment parameters including femoral I/E and tibiofemoral V/V angles, as well as tibial posterior slope were varied within the range defined according to practical surgical protocols and previous case reports. A parametric study was thus processed to numerically assess the effect of alignment parameter variability on postoperative tibiofemoral joint motion and contact mechanics.

3.2 Methods

3.2.1. Experimental Kinematic Study

In the current study, a force-controlled ISO 14243-1 gait simulation test was conducted using the Instron-Stanmore knee simulator. A commercial TKR device (Left knee, Triathlon®, Stryker Orthopaedics, Mahwah, NJ) was selected and evaluated during normal gait cycles derived by averaged knee joint loads calculated based on previous experimental and theoretical observation [107]. As shown in Figure 3.1, the experimental inputs included quantitative waveforms of 0-60° femoral flexion (rotational displacement), tibial I/E torque, tibial anterior-posterior (A/P) force and tibiofemoral axial compressive loading. In-vivo knee capsular and ligamentous restraints were simulated using soft pre-compressed springs, with linear constraining coefficient of 20N/mm and axial rotational restraint of 0.28Nm/degree [107].
Figure 3.1 Waveform inputs of femoral flexion angle (a), tibial internal (+)-external (-) torque (b) and anterior (+)-posterior (-) translational force (c), and tibial axial compressive loading (d) for the Instron-Stanmore knee simulator.

The femoral component was mounted on the simulator using a fixing phenolic block (Figure 3.2), using with a sagittal profile equivalent to the femoral condyle after the TKR surgical procedure. Geometry of the phenolic block was specified aligning with the standard anatomical references, and the femoral flexion was defined as uniaxial rotation with respect to a single axis within the phenolic block based on implant’s sagittal profile. Therefore, this instantaneous motion axis is the equivalent representation of the transepicondylar axis in the realistic TKR surgical protocol, and the specific shape of the phenolic block also guaranteed equivalent parallel cut for posterior femoral condyles with respect to the transepicondylar axis, and this direction possess an included angle of 3° with the femoral posterior condylar axis [85]. The tibial component, on the other hand,
was aligned and positioned within the resting cup according to the position of the femoral component. By default, the standard experimental setting assumed 0 mm of medial-lateral shift, 0° of tibial posterior slope, and 0° of relative I/E rotation between the femoral and tibial components, hence, the lowest articulating point of both femoral condyles should be coincident with the corresponding points at the tibial plateau at the stage of full extension. In addition, at the coronal plane, both components were axially aligned, indicating 0° of V/V rotation, which consistently corresponds to the ideal surgical condition: the tibial and femoral mechanical axes are collinear, while the angular difference between two anatomical axes is 5°.

Figure 3.2 CAD geometry of the phenolic block for fixing the TKR femoral component.

The experimental kinematic testing was then initiated by the aforementioned active loading through the actuation system of the Instron-Stanmore knee simulator. 50% bovine serum was used as joint lubricant. Motion data for 15 complete walking cycles at 0.8 Hz were sampled and collected for average.
3.2.2. Standard FE Model Establishment and Experimental Validation

Based on the experimental conditions described above, a 3D computational finite element (FE) model was developed to numerically predict the postoperative TKR kinematics and tibiofemoral contact mechanics. Geometries of the femoral fixing bracket and phenolic block, as well as the tibial resting cup, were built in SolidWorks (v2012, SolidWorks Corp., Waltham, MA), while CAD geometry of the TKR device, was captured and digitally converted using 3D scanner with multi-laser precision (NextEngine Inc., Santa Monica, CA). The complete geometrical assembly was then further processed and cleaned in HyperMesh (v11.0, Altair Engineering Inc., Troy, MI). The components were aligned based on experimental cases, and initially placed in the 3D modeling space according to the unloaded relative positions from the Instron-Stanmore knee simulator testing. Surface-based rigid ties were created to fasten different components within tibial or femoral assembly and avoid relative motion. Brick solid FE elements were employed to mesh the femoral bracket and phenolic block, while other components were numerically discretized using 3D tetrahedron elements in HyperMesh. The linear and rotational kinematic constraints were imposed in the FE model through parallel springs connected to the tibial cup, with equivalent stiffness constants with the machine settings.

In the current study, the TKR device consists of 3 components, the femoral, tibial components and tibial insert, of which the femoral and tibial components are made of CoCr and Titanium alloys, respectively. Material data (Table 3.1) for ASTM F75 and F136 were used to quantify the corresponding metallic material behaviors (classical linear elasticity, isotropy). The tibial insert was made of UHMWPE, of which the detailed
material property was characterized utilizing published mechanical testing data [108, 109]. It is indicated that UHMWPE presents linear elasticity, nearly incompressible elastic \((E=876\text{MPa}, \nu=0.46)\) and rate-dependent plastic deformation under monotonic tensile/compressive loading, which can be mathematically described by nonlinear elastoplastic material representation following von-Mises yield criterion and isotropic straining hardening [110-112]. Standard testing data (true stress-strain curves) obtained at multiple loading rates from literatures were employed for material modeling and further predict the rate-dependent response via interpolation based on inputs. Additionally, bone cement, modeled as a solid block \((E=3400\text{MPa}, \nu=0.3 [113])\), was modeled into the cavity of the tibial cup and retained the same level of depth as in the experimental testing to preserve the overall inertia effect during tibiofemoral motion.

Table 3.1 Material parameters for defining the CoCr and Ti alloys

<table>
<thead>
<tr>
<th>ASTM F75 (Co-28Cr-6Mo)</th>
<th>Young’s Modulus (Gpa)</th>
<th>Poisson’s Ratio</th>
<th>Density (g/cm(^3))</th>
</tr>
</thead>
<tbody>
<tr>
<td>210</td>
<td>0.3</td>
<td>8.3</td>
<td></td>
</tr>
<tr>
<td>ASTM F136 (Ti-6Al-4V)</td>
<td>115</td>
<td>0.33</td>
<td>4.4</td>
</tr>
</tbody>
</table>

As Figure 3.1 shows, waveform inputs of the experimental knee simulator testing, i.e. the femoral flexion angle, tibial I/E torque and A/P translational force, and tibiofemoral compressive loading, were equally applied in the FE model as amplitude-based boundary conditions. The femoral bracket and phenolic block were defined as rigid bodies with respect to the middle point of the femoral flexion axis, and the tibial cup was rigidized as well with respect to the central point of the connecting line between two
lowest articular contacting points. Therefore, the prescribed motion/loading conditions were specified at the rigid body reference points for both femoral and tibial assemblies. The joint axial compressive loading was split 60/40 at the medial and lateral condyle, to reproduce the identical experimental joint force condition and match the realistic physiological joint loading distribution.

Figure 3.3 (a) Finite element model of the Instron-Stanmore simulator testing components and (b) side view of the femoral bracket and top view of the tibial assembly during the corresponding experimental testing.

Within the current FE model (Figure 3.3), same as the experimental set-up, 50% bovine serum was involved as lubricant for joint contact. Identified as typical non-Newtonian fluids, the viscous nature of bovine serum was mechanically described using Power-law, which was commonly applied in modeling nonlinear deviatoric fluid and featured in most commercial FE solvers. As the following equation shows,

\[
\eta = k\dot{\gamma}^{n-1} (\eta_{\text{min}} \leq \eta \leq \eta_{\text{max}})
\]  

(3.1) [114]
$k$ and $n$ represent the flow consistency index and flow behavior index, respectively, $\dot{\gamma}$ is the shear rate, and $\eta$ is the apparent viscosity. According to published experimental measurement about lubricating film [115], bovine serum’s viscosity decreases along with increasing shear rate, numerically, showing $n < 1$ in Equation 3.1, and the detailed $\eta - \dot{\gamma}$ curve was fitted to determine constants $k$ and $n$. Meanwhile, the incompressible volumetric response of the lubricant was governed by linear Hugoniot Equation of State (Equation 3.2).

$$U_s = c_0 + sU_p$$

(3.2) [114]

where $U_s$ and $U_p$ represent linear shock velocity and particle velocity, respectively. $c_0$ and $s$ are material specific constants for bovine serum, with $c_0^2 \rho$ is equal to the elastic bulk modulus at small nominal strain. The combined solid-fluid modeling system was expected to maintain quasi-static state during the loading cycle, therefore, weak and stable shock was defined for fluidic behavior, and the linear shock velocity $U_s$ was set to be constant with $s = 0$. Initial fluid velocity was then defined as initial conditions for the FE model, and the initial volumetric occupation of fluid was determined by identifying cavity of the tibial cup which was filled by lubricant within the predefined fluid domain. Surface contact with penalty-based algorithm to prevent interception and penetration was defined between the tibiofemoral articular surfaces and biphasic interface between fluidic lubricant and solid components. Coupled Eulerian-Lagrangian (CEL) dynamic FE analysis was carried out afterwards using Abaqus/Explicit (v6.11, Simulia Corp., Providence, RI). Simulation results were then compared with the experimental outputs, to
evaluate the statistical equivalence and validate the computing accuracy/robustness of the FE model.

3.2.3. Post-processing Case Study for Parametric Analysis

Once the standard FE modeling calculations were validated by the corresponding experimental results, a series of FE modeling studies and a full-factorial design of experiment (DOE) study were conducted to predict potential biomechanical outcomes resulted from TKR postoperative malalignment. Referring to alignment variation in surgical cases observed from previous studies [92, 94-96], in the current study, for alignment femoral I/E rotation angle was set to vary from -6° to 6° (external +, internal -), and tibiofemoral V/V ranged from -5° to 5° (varus +, valgus -), while tibial posterior slope varied in-between 0° and 7°. Variability of all three parameters was deemed to be continuous thus numerous combinations can be obtained.

Table 3.2 Cases selected for further FE simulation

<table>
<thead>
<tr>
<th>Case Labels</th>
<th>Femoral I/E Angle (degree)</th>
<th>Tibiofemoral V/V Angle (degree)</th>
<th>Tibial Posterior Slope (degree)</th>
</tr>
</thead>
<tbody>
<tr>
<td>ExR2</td>
<td>2</td>
<td>0</td>
<td>0</td>
</tr>
<tr>
<td>ExR4</td>
<td>4</td>
<td>0</td>
<td>0</td>
</tr>
<tr>
<td>ExR6</td>
<td>6</td>
<td>0</td>
<td>0</td>
</tr>
<tr>
<td>InR2</td>
<td>-2</td>
<td>0</td>
<td>0</td>
</tr>
<tr>
<td>InR4</td>
<td>-4</td>
<td>0</td>
<td>0</td>
</tr>
<tr>
<td>InR6</td>
<td>-6</td>
<td>0</td>
<td>0</td>
</tr>
<tr>
<td>TL3</td>
<td>0</td>
<td>0</td>
<td>3</td>
</tr>
<tr>
<td>TL7</td>
<td>0</td>
<td>0</td>
<td>7</td>
</tr>
<tr>
<td>Var3</td>
<td>0</td>
<td>3</td>
<td>0</td>
</tr>
</tbody>
</table>
Twelve representative cases were selected and analyzed individually based on the standard model and updated alignment conditions in Abaqus, of which the brief information and case labels are listed in Table 3.2. The DOE study was further conducted and the results were fitted to a higher-order mathematical function (“Response Surface”) with satisfying the model observations from the 12 selected cases. Considering the clinical significance, peak tibiofemoral contact pressure was defined as the main biomechanical response, and the DOE procedure was performed in HyperStudy (v11.0, Altair Engineering Inc., Troy, MI).

<table>
<thead>
<tr>
<th>Var5</th>
<th>0</th>
<th>5</th>
<th>0</th>
</tr>
</thead>
<tbody>
<tr>
<td>Val3</td>
<td>0</td>
<td>-3</td>
<td>0</td>
</tr>
<tr>
<td>Val5</td>
<td>0</td>
<td>-5</td>
<td>0</td>
</tr>
</tbody>
</table>

Figure 3.4 Representative modeling cases for varied alignment DOE study (ExR6, InR6, TL7, Val5 and Var5)
3.3 Results

3.3.1 Results of the Experimental Kinematic Study and the Standard FE Model

The predicted postoperative tibiofemoral kinematics after TKR, including the tibial A/P translation and I/E rotation during a full gait cycle, was exported from both the experimental Instron-Stanmore testing and the standard FE model. As shown in Figures 3.5a and b, experimental and simulation outputs demonstrate good agreement and convergence in terms of trend and magnitudes during a full gait cycle. In Figure 3.5a, positive angular displacement refers to external tibial rotation, while in Figure 3.5b, positive linear displacement refers to anterior tibial translation. The tibial internal-external rotation reported by the experimental testing ranged from $3.19^\circ$ to $-8.50^\circ$, while the model-predicted results varied within $3.22^\circ$ and $-7.97^\circ$ interval, therefore, the model prediction was $0.94\%$ and $6.24\%$ different from the experimental observation for peak tibial internal and external rotation, respectively. Similarly, the tibial anterior-posterior translation recorded in the experimental testing ranged from 2.21 mm to -3.51 mm, while the corresponding range calculated from the FE model was between 2.38 mm and -2.82 mm, so the model prediction was different from the experimental outcome by $7.69\%$ and $19.66\%$ for peak tibial anterior and posterior translation, respectively. Based on peak motion magnitudes, the standard FE model was able to closely approximate the experimental tibial I/E rotational kinematics, yet tended to slightly overestimate the peak tibial posterior, and underestimate the tibial anterior translation. To compare the overall difference between the variance of the two sets of data, a multivariate analysis of variance (MANOVA) was conducted in SAS (v9.0, SAS Institute Inc., Cary, NC). The
statistical analysis revealed that the difference between the experimental and model results was insignificant, with calculated $p$ value of 0.80.

![Comparative Tibial I/E Rotational Displacement](image1)

![Comparative Tibial A/P Translational Displacement](image2)

Figure 3.5 Comparative curves of tibiofemoral translational (a) and rotational (b) motion exported from experimental knee simulator and the standard FE model
Figure 3.6 (a) Variation of peak tibiofemoral contact pressure within lateral and medial condyles and (b) total contact area during the gait cycle.

Peak tibiofemoral joint contact pressure across both lateral and medial condyles during the full gait cycle was also exported from the FE model (Figure 3.6a) (for simplification, “tibiofemoral contact pressure” refers to the peak value within the entire contact area in contexts). The varied contact pressure magnitude peaked at approximately
5%, 40% and 55% of the gait cycle, and the specific values were 20.95, 18.94 and 20.82 MPa. Physically, these three peak pressures could be associated with phases of heel strike (initial contact), heel off (terminal stance) and pre-swing during the gait cycle [116], at femoral flexion angles of approximately 5°, less than 5°, and 30°, respectively. On the other hand, the total tibiofemoral contact area variation obtained through the FE modeling study presented local peaks at 5% (103.75 mm²), 15% (132.62 mm²), 35% (137.87 mm²) and 45% (135.74 mm²) of the gait cycle (Figure 3.6b), with the global maximum contact area occurring at 35% of the entire process, which related to a transitional state from the phase of mid-stance to heel off.

Figure 3.7 illustrates the tibiofemoral contact pressure contours when either contact pressure or area reached local maximum during the gait cycle. Within each time frame, relatively higher pressure magnitude and contact area can be found within the medial condyle, which is attributed to asymmetric axial loading distribution across both medial and lateral compartments. Tibial I/E rotation relative to femoral condyles during the motion cycle affected the specific tibiofemoral contact points at both medial and lateral sides: at the initial stage (approximately 0-25%) of the gait cycle, when tibial component presented notable external rotation, the contact area at the medial side was offset closely to the posterior end of the tibial plateau. Similarly, the contact area at the lateral side was offset with respect to the tibial plateau boundary when tibial internal rotation occurred.
Figure 3.7 Tibiofemoral contact pressure contours at (a) 5%, (b) 15%, (c) 35%, (d) 40%, (e) 45% and (f) 55% of the gait cycle, presenting local maximal magnitudes of contact pressure and contact area.
3.3.2 Results of the Parametric Malalignment Case Study and Design of Experiment

On the basis of the validated standard FE model, results regarding tibial kinematics and tibiofemoral contact pressure were collected from the 12 representative models with postoperative TKR alignment variation for parametric study. As depicted in Figure 3.8, the output curve pattern of tibial internal-external rotation for cases with various femoral I/E, tibiofemoral V/V angles or tibial slope was generally consistent. Similar curve trend was also found in the exported curves of tibial anterior-posterior translation (Figure 3.9). Specifically, Figure 3.8a reveals that in the standard case, the average curve value of the tibial internal-external rotation along the full motion cycle was -1.49±2.95°, while the mean value increased by 23.26% (0.35°), 40.19% (0.60°), and 62.15% (0.93°) in the cases with 2° (case label: ExR2), 4° (case label: ExR4) and 6° (case label: ExR6) femoral external rotational angle, respectively. On contrary, for cases with 2° (case label: InR2), 4° (case label: InR4) and 6° (case label: InR6) internally rotated femoral component, the averaged curve value of the tibial internal-external rotation decreased by 36% (0.54°), 56.36% (0.84°) and 68.73% (1.03°), respectively. These aforementioned findings indicate that increased femoral external rotation in alignment could in general lead to higher tibial external but lower internal rotation, while increased femoral internal rotation, conversely, reduce tibial external but raise internal rotation during gait.

Figure 3.8b presents the effect of postoperative tibial posterior slope on the tibial internal-external rotation. Significant change was noticed in the case with 7° tibial slope (case label: TL7), of which the overall average was 0.70±3.08°. The peak tibial external
rotation was 7.70°, approximately 2.4 times of that in the standard case (3.22°), while the highest internal rotation was 5.37°, which reduced by 32.59% (2.60°) compared with the standard case (7.97°). The case with 3° tibial slope (case label: TL3) showed intermediate curve offset. The average value increased by 39.09% (0.58°) relative to the case without tibial inclination, and lesser peak tibial external and internal rotation were observed along the motion cycle.

As Figure 3.8c shows, the effect of tibiofemoral V/V angle can be identified in 3 different time intervals. With respect to the standard case, at the initial stage (between 0 and approximately 25%) of the gait cycle, 3° (case label: Var3) and 5° (case label: Var5) varus alignment demonstrated higher tibial external rotation, with peak magnitude increased by 24.9% (0.80°, Standard: 3.22°) and 66.99% (2.16°), respectively; at the middle stage (between 25 and 62%), varus angles led to intermediate reduction of the maximum tibial internal rotation by 12.45% (0.99°, Var3; Standard: 7.97°) and 15.18% (1.21°, Var5); at the final stage (beyond 62%) of the motion cycle, slight tibial internal rotation (<0.1°) was recorded from both cases, close to the standard case prediction.

More evident alteration was observed in the cases with 3° (case label: Val3) and 5° (case label: Val5) valgus alignment. In both cases, at the initial stage, calculated peak tibial external rotation reduced by 15.94% (0.51°, Val3) and 8.75% (0.28°, Val5); at the middle stage of gait cycle, tibial internal rotation increased considerably by 29.33% (2.33°, Val3) and 21.18% (1.69°, Val5); at the final stage, valgus-aligned cases presented moderate tibial external rotation (≈2°).
Figure 3.8 Comparative case outputs of the tibial internal-external rotation variation during the gait cycle with different postoperative TKR alignment: (a) TKR aligned with additional I/E angle; (b) TKR aligned with additional tibial slope; (c) TKR aligned with additional V/V angle.

Figure 3.9a shows minimal influence of femoral I/E alignment on the tibial anterior-posterior translation, with maximum difference of 0.5 mm in peak magnitudes from various cases compared with the standard case. Figure 3.9b shows the altered response pertaining to the tibial posterior slope. Evident curve offset towards anterior translation has been noted, and the averaged tibial displacements in TL3 and TL7 cases were 17.54 and 40.34 times of that in the standard case (0.05±1.52 mm). Especially, in the case with 7° tibial slope, the maximum tibial anterior translation was calculated to be 6.21 mm, which was 3.84 mm higher than the standard case. This figure straightly indicates increased postoperative tibial posterior slope could greatly enhance tibial anterior, while lessening the posterior translation.
From Figure 3.9c, slight difference among different curves can be found in a single time interval. Between approximately 15% and 60% of the gait cycle, cases with extra tibiofemoral varus angles showed slight higher average curve values (Var 3: -0.55±1.81 mm, Var 5: -0.51±1.77 mm) than the standard case (-0.66±1.75 mm), whereas valgus-aligned cases slightly increased the tendency of posterior movement by showing lower curve average (Val 3: -0.97±1.80 mm, Val 5: -1.16±1.79 mm).

Regarding the variation of the tibiofemoral contact pressure during the gait cycle, the 12 representative cases showed remarkable variation among each other. Specifically, as Figure 3.10a shows, for tibiofemoral contact pressure, relative to the standard condition, the case with 2° femoral external rotation showed local peaks at 5%, 15%, 40% and 50% of the entire motion cycle, and the values were 14.31, 16.06, 15.11 and 17.80 MPa, respectively. The local maximums occurred at time points close to but not the same with those in the standard case. Case ExR 4 and ExR 6 demonstrated similar curve trend with each other, and for both models, the tibiofemoral contact pressure reached major local peaks at 15% (ExR 4: 19.42 MPa, ExR 6: 19.02 MPa), 40% (ExR 4: 16.32 MPa, ExR 6: 18.35 MPa) and 50% (ExR 4: 15.42 MPa, ExR 6: 18.21 MPa). Meanwhile, the global maximum tibiofemoral contact pressure can be found at 15% of the gait cycle in both cases, with slightly lower magnitude than the standard case.
Figure 3.9 Comparative case outputs of the tibial anterior-posterior translation variation during the gait cycle with different postoperative TKR alignment: (a) TKR aligned with additional I/E angle; (b) TKR aligned with additional tibial slope; (c) TKR aligned with additional V/V angle

More distinct and random curve patterns were observed in the cases with extra femoral internal rotational alignment angles (Figure 3.10b). Case InR2 showed local peaks at 5%, 30%, 40% and 55% of the gait cycle, and the specific pressure values were 17.91, 16.62, 19.27 and 20.44 MPa, respectively. Major local maximums in Case InR4 occurred at 10% (18.01 MPa), 30% (18.17 MPa), 50% (17.32 MPa) and 70% (13.96 MPa), and Case InR6 had partially parallel curve variation, in which higher tibiofemoral pressure magnitudes were achieved at 10% (16.93 MPa), 50% (16.18 MPa) and 70% (17.18 MPa) of the motion cycle. Overall, it can be quantitatively noticed that increased femoral internal rotational alignment could be associated with decreased overall global peaks.
Higher peak tibiofemoral contact pressure (26.25 MPa) was predicted in the case with 3° tibial posterior slope, which increased by 26.08% with respect to the magnitude recorded at the same stage of gait in the standard case. Local maximums were noticed at time spots of 15% (20.95 MPa), 45% (17.12 MPa) and 55% (26.25 MPa), whereas in the case with 7° tibial posterior slope, local maximums were found at 15% (14.66 MPa) and 45% (17.49 MPa), with significantly reduced peak pressure values during the motion cycle (Figure 3.10c).

Figure 3.10d and e reveal the influence of tibiofemoral V/V rotational angle. Excessive varus or valgus alignment led to enlarged tibiofemoral contact pressure, and in terms of calculated values, cases with 5° varus/valgus rotation in general presented higher curve level than cases with 3° varus/valgus rotation. Specifically, Figure 3.10d shows major local peaks of tibiofemoral contact pressure at 5% (18.30 MPa), 15% (25.57 MPa) and 50% (22.93 MPa) in the case Val3, while in the case Val5, curve peaked at 5% (26.03 MPa), 15% (26.33 MPa), 35% (25.54 MPa) and 50% (24.09 MPa), which shows relatively higher pressure magnitudes same time points. As Figure 3.10e demonstrates, case Var3 and Var5 reached local maximums at similar time points, which can be noticed between Val3 and Val5 as well. For the case with 3° varus alignment angle, the overall highest tibiofemoral contact pressure was obtained at 45% (26.23 MPa), and presented additional local peak at 15% (22.52 MPa) of the entire gait cycle. Case Var5 revealed higher global peak pressure value (27.84 MPa) at 15% of the motion cycle, with another local maximum occurred at time point of 50% (24.92 MPa).
Figure 3.10 Comparative case outputs of the peak tibiofemoral contact pressure variation during the gait cycle with different postoperative TKR alignment: (a)-(b) TKR aligned with additional I/E angle; (c) TKR aligned with additional tibial slope; (d)-(e) TKR aligned with additional V/V angle

Numerical contribution of all 3 alignment variables towards the major functional response, i.e. the peak tibiofemoral contact pressure magnitude during the gait cycle, was assessed utilizing DOE, and fitted by existing calculation results from the
representative 12 cases. As Figure 3.11 shows, 3D response surfaces were established with continuously varied tibiofemoral V/V and femoral I/E rotational alignment angles as X and Y coordinates, and created in series for increasing posterior slope angles with increment of 1°. All 3 variables were prescribed within the pre-defined range, and good statistical correlation was found between the existing case results and the response surface derived by DOE ($R^2 = 0.93$). Within the saddle-shaped response surface at all tibial slope levels, two major estimations can be expected: (a) as the femoral I/E alignment angle approaches the neutral state, the predicted peak tibiofemoral contact pressure tends to be higher; (b) as the tibiofemoral V/V alignment angle approaches zero degree, the predicted peak tibiofemoral contact pressure tends to be lower. In addition, by comparing response surfaces at multiple tibial slope levels, the “saddle” is lifted up as tibial slope varies towards to the middle of its value range.

3.4 Discussion

The current study utilized combined investigations of force-controlled Instron-Stanmore knee simulator testing and dynamic FEA simulation to assess the effect of various alignment variables in determining the tibiofemoral motion and contact mechanics. The experimental knee simulation testing was directly implemented to simulate the common postoperative daily walking activity. Compared with in-vivo or in-vitro cadaveric testing, the Instron-Stanmore knee simulator provides great controllability and flexibility in measurement regarding testing inputs [71], and is capable to predict accurate gait motion with consistent repeatability and reproducibility [77, 108].
Figure 3.11 Response surface obtained on the basis of post-processing DOE study and existing case study results at tibial posterior slope of (a) 1°, (b) 3°, (c) 5° and (d) 7°, indicating the influence of all 3 major TKR rotational alignment parameters on the maximum tibiofemoral contact pressure magnitude.

Additionally, the FEA models employed in the current study introduced lubricant domain, which matched the experimental setting to simulate the in-vivo synovial fluid for joint lubrication. The function of synovial fluid includes lubricating the articular surfaces, damping and dissipating impact loading to facilitate joint movement. Therefore, the current FEA models can be expected to predict more postoperative joint kinematics relative to realistic scenarios.
The statistical evaluation between the Instron-Stanmore experimental testing and parallel simulation results revealed good convergence regarding postoperative tibiofemoral kinematics, including tibial anterior-posterior translation and internal-external rotation, which proves the validity of the FEA model. Minor gaps between the experimental and computational curves observed in Figure 3.5 could be associated with the calculating perturbation during the computational simulation. Also, in the actual experimental testing, stiffness of the restraining spring and viscosity of the bovine serum could also present variance and discrepancy compared with the theoretical value inputs in the FEA model. In addition, residual inertial effect existed in the current quasi-static FEA model could slightly affect the final computational prediction as well.

For various cases with different rotational alignments presented in the parametric malalignment case study, inputs of tibial internal-external torque and anterior-posterior force remained consistent among cases. Kinematic difference in tibial I/E rotation and A/P translation could be quantitatively associated with the axial joint loading, asymmetrical geometries of the TKR femoral condyles and tibiofemoral interaction. Theoretically, since approximately 60% of the total joint compressive loading is distributed at the medial tibial condyle, and the medial femoral condyle is geometrically bigger than the lateral, higher magnitudes of tibiofemoral contact pressure and contact area can be anticipated at the medial tibial condyle, as Figure 3.7 demonstrates. Due to relative movement between tibial and femoral components, friction is originated from contact pressure along the contacting area (i.e. total contact force) and exerted onto both tibial condyles, while higher force is generated at the medial side, driving tibial
component to translate anteriorly and rotate externally once femoral flexion increases. Similarly, tibial assembly is driven to rotate internally when flexion decreases. In the current study, when femoral external or internal rotation (femoral I/E alignment angle) in alignment was present, corresponding to ExR or InR cases, shift of tibial I/E rotation curve to positive (external) or negative (internal) direction, respectively, could be expected due to the effect of the tibiofemoral friction. Consistent observation was also noted and addressed from experimental study conducted by Haider et al utilizing the same Instron-Stanmore knee simulator system [71] for 7 different TKR devices at different alignment setups. The tibial internal-external rotation variation curve was shifted towards to the same direction as the preexisting rotation because of the fact that predicted tibial rotation was measured based on the original neutral reference, which could well justify the curve pattern in Figure 3.8a in the current study as well. On the other hand, from the same study, quantitative change in the tibial anterior-posterior translational displacement could be neglected [71]. Similarly, little curve difference can be found in Figure 3.9a in the current study.

Giffin et al. conducted a cadaveric study addressing the effect of the tibial slope on knee kinematics and cruciate ligaments tensioning [117]. According to the finding, due to the posterior tibial slope, the axial joint compressive loading can be decomposed into two components: the normal and the tangential force components, with the latter pointing to the anterior direction. Hence, it is expected that the tibial component would demonstrate extra anterior translational displacement, which explains the observation found in Figure 3.9b: as tibial posterior slope increases, more displacement in the anterior
direction can be expected. This prediction is also supported by previous studies [118, 119]. Furthermore, from a clinical perspective, it is widely accepted that excessive tibial anterior displacement is beneficial in increasing the moment arm of the extensor mechanism, smoothening the deep flexion motion and resulting in a higher range of motion [93]. The anterior force difference at both medial and lateral tibial condyles could be significantly increased due to the introduction of tangential component of the contact force, which is also asymmetrically distributed (medially biased), causing extra tibial external rotation and contradicting internal rotation. As a result, shift of the tibial internal-external variation curve towards to the external direction is expected as shown in Figure 3.8b.

In aforementioned study conducted by Haider et al., 10° varus and valgus angles were incorporated in the varied tibiofemoral rotational alignment. However, the authors did not report a significant difference regarding tibial motion, since the action line of joint axial loading remained passing through the connecting line between contacting points at both tibial condyles, avoiding joint inclination and laxity change [71]. In the current study, a minor variance was observed regarding the tibial anterior-posterior translation in the cases with 3° or 5° preexisting V/V alignment. Comparatively, for calculated tibial internal-external rotation, Var3 and Var5 showed intermediate difference, while Val3 and Val5 revealed substantial alteration. Extra varus or valgus angle could increase the tendency of joint loading concentration. In general, this will result in reduced joint space on one side and increased space on the other, equivalently increasing rigidity to relative displacement in one direction but easing the motion in the
other direction. Consequently, in cases Var3 and Var5, tibiofemoral contact pressure at the medial side got enlarged, which prevented tibial internal rotation. As the incremental varus alignment angle accumulated, more rotational resistance recruited accordingly. In contrast, in cases Val3 and Val5, tibiofemoral contact pressure was concentrated at the lateral tibial condyle, which partially counterbalanced the medially-biased joint axial compressive loading and laterally translated the action line. In this way, tibial internal rotation was increased extensively due to not only release of the medial joint space, but also reduction of counterbalancing moment originated from the joint compression for the experimental tibial internal-external torque input.

Under most circumstances, excessive postoperative tibial component rotation and translation may cause patellar instability, soft-tissue impingement, pain and knee joint malfunction [85], which further emphasizes the importance of correct TKR alignment. Even though extensive tibial anterior translational displacement is sometimes desirable for certain patient groups to perform deep flexion/squatting motion, according to Bellemans et al. [93] and Wasielewski et al. [120], anterior subluxation may bring about undesirable wear problem associated with the posterior tibial tray edge.

Liau et al. conducted combined studies utilizing in-vitro experiment and computational simulation to explore the effect of tibiofemoral alignment on knee joint contact pressure [70, 121]. Knees with 1°, 3° and 5° preexisting femoral internal alignment showed slightly lower peak contact pressure, while specimens with 1°, 3° and 5° tibiofemoral varus tilt showed tremendous value increase. In the current study, similar conclusion can be made based on predictions from InR and Var cases. As for Var cases,
Val cases also lead to pressure concentration at the lateral tibial condyle. Given the fact that the knee joint axial loading was initially medially-shifted, the global peak tibiofemoral contact pressure values during postoperative gait cycle at the lateral compartment was expectedly smaller than those calculated from corresponding Var cases. Interestingly, the 3° tibial posterior slope case showed significantly higher peak contact pressure value than the standard case, while case with 7° tibial slope in general presented lower contact pressure level during the motion cycle. This could be associated with the variation in geometrical congruence across tibial plateau, since for case TL7, the resting site for the femoral component was more posterior than that for TL3.

Overall, pressure variation and peaks were highly dependent on geometrical conformity and relative position/motion between tibial and femoral TKR components. Especially, several local pressure peaks during the motion cycle could be associated with edge contact related with tibial malrotation attributed to varied rotational alignment, which could risk the implant components in regular use [70]. Meanwhile, close relation between contact pressure magnitude and wear rate has been well recognized [70, 105, 106]. Relative tibial translational and rotational displacement (travelled distance) have been linked with wear rate as well in previous studies [103, 122]. As a result, given the numerical influence of varied alignment on contact pressure and postoperative kinematics, tibiofemoral rotational alignment significantly determines the long-term performance of the TKR device.

The DOE parametric study served as a potential reference tool for determining appropriate tibiofemoral rotational alignment. As shown in Figure 3.11, this analysis
utilized high order mathematics function for prediction. The surface adaptive response demonstrated good correlation with the results obtained from existing parametric cases, showing reliable prediction provided by DOE. However, the current method only involved peak tibiofemoral contact pressure during gait as the performance indicator for functional assessment. In order to comprehensively determine the optimum rotational alignment parameters based on realistic clinical condition, postoperative kinematics, tibiofemoral contact area, soft tissue (e.g. posterior cruciate ligament) tensioning, etc. could also be involved into the post-processing DOE parametric estimation. Also, 12 representative cases were employed in the current study to testify the statistical correlation, which could be insufficient in terms of sample size for validation. Ideally, more models with varied alignment conditions would be preferred to further calibrate the response surface for higher accuracy.

Another major limitation associated with the current study is the discrepancy between the experimental knee simulator testing as well as the parallel computational FEA model, and clinical practice. Even considered as the most common postoperative daily activity for TKR patients, gait motion only refers to 0-60° of knee flexion, which means the effect of varied rotational alignment on tibiofemoral kinematics and contact mechanics at deep knee flexion still remains unknown. Also, the Instron-Stanmore knee simulator focuses on solely theoretical replication for the knee joint, and kinematic constraints from soft tissue (muscular, tendinous and ligamentous structures) are greatly simplified, which could contribute to inaccuracy in predicted motion. Particularly, patella and patellofemoral joint are not included in the system. However, patellar tracking is
tightly related with the tibial internal-external rotation and anterior-posterior translation. Finally, as mentioned earlier, tibiofemoral contact relies heavily on the geometrical conformity of the tibial tray, and great variation in geometry exists among different implant designs. From this standpoint, to develop a higher level understanding, it would be recommended to involve implant designs with multiple levels of contact/geometrical conformity.

3.5 Conclusions

The current study established an analytical computational FEA platform validated experimentally using Instron-Stanmore knee simulator testing, and enabled post-processing parametric study, to quantitatively assess the biomechanical effect of three major rotational alignment parameters for the TKR device, femoral internal/external angle, tibiofemoral varus/valgus angle and tibial posterior slope, on postoperative tibiofemoral kinematics and contact pressure during common gait motion.

Majorly, extra femoral internal-external in alignment caused tibial rotation in the same direction. Tibial slope significantly led to increase of tibial external rotation and anterior translation. Valgus tibiofemoral alignment could be associated with increased tibial internal rotation especially at the middle phase of gait. Regarding the tibiofemoral contact pressure, additional femoral external rotation, higher tibial posterior slope and lower varus/valgus tibiofemoral inclination in alignment may contribute to reduced pressure magnitude.

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CHAPTER 4  INTRODUCTION OF EFFECT OF PATELLAR COMPONENT THICKNESS ON PATELLAR KINEMATICS AND PATELLOFEMORAL JOINT MECHANICS FOLLOWING TOTAL KNEE REPLACEMENT

As one of the most prevalent surgical procedures performed nowadays, total knee replacement (TKR) has gained more and more attention. Over 500,000 TKR surgeries are performed each year in the United States, and it is estimated that this number will soon increase to over 3.48 million per year due to the aging baby-boomers, more common obesity problem and the trend of knee reconstruction for both younger and older patients [1]. Also, total joint replacement surgery is allowed in several cases for elderly patients with co-morbid conditions to benefit their day-to-day activities [2]. Therefore, increasing research interests and investment have been made regarding the implant longevity, failure mechanisms, biocompatibility and design optimization, which are some major concerns associated with TKR.

The patellofemoral (PF) joint, consisting of the patella and femoral trochlea, is critical to the knee joint stability primarily through its role in the extensor mechanism. The patella displaces the patellar tendon further away from the tibiofemoral contact sites during knee flexion/extension, and increases the mechanical advantage of the extensor mechanism by transmitting the extending force at a greater distance from the knee flexion center by introducing moment arm for the patellar tendon [18, 19]. According to previous
observations, the patella reduces the quadriceps force required in knee extension motion by 15% to 30% by increasing the moment arm relative to the knee flexion axis [16-18]. The magnitude of the moment arm varies during the range of motion as a result of the patellar kinematics. Simon et al. concluded that the lever arm of the quadriceps increases by about 10% at full flexion, then increases to 30% by 35°-45° from full extension, and then decreases as the knee approaches to full extension [123].

Even remaining as a controversial surgical procedure, patellar resurfacing has been widely adopted in TKR to eliminate postoperative anterior knee pain, specifically for indications such as patellar inflammatory arthritis or greater patellar arthritic changes in osteoarthritic patients [124]. Ideally, the natural patellar thickness should remain unchanged after the patellar component implantation procedure to maintain the efficiency of the quadriceps extensor mechanism. Therefore, the thickness of patellar bone resection should be equal to the artificial patellar component [25].

Even though it is generally assumed that the restoration of preoperative patellar thickness is the most desirable outcome of the patellar resurfacing procedure in TKR surgery [124, 125] specific physical conditions could vary significantly among patient groups. A specific example refers to the variation of intact patellar thickness among races. An average natural patellar thickness of 23-30 mm was reported for Western population as compared to 21-22 mm in East Asian population (Chinese and Korean), and 20–23 mm in Southeast Asian population (Malaysian) [126]. Thinner natural patella present a challenge while following current implantation protocols recommending that residual patellar bone thickness should not be less than 12-15 mm after patellar
resurfacing procedure. Additionally, other factors such as specific TKR design features, and clinical practitioners’ experience could also result in suboptimal postoperative patellar thickness levels.

Clinical studies identified that anterior knee pain and dysfunction as one of the major complications after TKR surgery, and specifically residual patellofemoral pain exists in 5-45% of TKR patients [127-129]. Previous studies have utilized in-vivo and in-vitro methods to address the influence of patellar thickness on anterior knee pain from a knee biomechanics perspective.

It has been well-recognized that the physiological knee joint function is prone to be affected by the patellar thickness change. Dennis et al. reported that the average peak flexion angle ranged from 100° to 110° after patellar resurfacing [130], which was less than the normal range (120° or above). Similarly, Bengs et al. designed an intraoperative study to evaluate the effect of patellar thickness on knee flexion during TKR surgery using 4 augmented patellar trials for 30 testing subjects (21 females, 9 males). Final statistical data revealed that passive knee flexion decreased 3° for every 2-mm increment of patellar thickness [131]. Such observation could attribute to an “overstuffed” patellofemoral joint after patellar resurfacing procedure. Overstuffing typically refers to unreasonably thick patella which may result from various factors associated with design and surgical inaccuracy, including inadequate bone cut, excessively thick UHMWPE patellar component, excessive thickness of residual trochlea groove and trochlear metal flange. Previous studies summarized that patellofemoral joint overstuffing leads to anterior knee pain, decreased range of motion (ROM), patellofemoral maltracking,
increased patello-femoral joint compressive and shear force, and excessive wear [131-133]. Specifically, decreased ROM mentioned above could be ascribed to the fact that a thicker patella increases the arc which extensor mechanism must travel to accomplish normal knee joint motion, introducing unnecessary kinematic hindrance thereby decreases passive knee flexion [134].

On the other hand, reduced postoperative patellar thickness is usually associated with patellar over-resection, which was noted to correlate with anterior patellar over-strain in previous studies. Through tracking dynamic flexion for 10 fresh, paired human cadavers, Reuben et al. recorded significantly higher strain in the specimens which had less than 15 mm of residual bony patella, and ultimately less than 25 mm of composite patellar thickness [135]. This conclusion is consistent with work conducted by Wulff et al., Yoo et al. and Jujo et al., who also identified that an over-resected patella is susceptible to bony fracture [133, 136, 137]. Fitzpatrick et al. further revealed that among specimens with multiple thickness levels, the thinnest patella presented highest peak strain, since greater bone volume enables patella to dissipate load to avoid stress concentration [138].

Furthermore, Cheng et al. measured in-vivo patellar tracking kinematics for 6 TKR subjects (3 male and 3 female) during stair climbing using fluoroscopy, and concluded that in general the quadriceps force increases with thinned and decreases with thickened patella. Also thinner patella could cause reduced patella-femoral (PF) joint reaction force and vice versa [139]. Using similar fluoroscopy technique on 81 subjects, Stiehl et al. found that increased patellar thickness is beneficial at 35° flexion but may not
affect the lever arm significantly at the full extension state [140], suggesting that increased patellar thickness correlates with reduced quadriceps loading especially at intermediate flexion, while such correlation may not be consistent at higher flexion angles.

In summary, although great effort has been made to assess the effect of patellar thickness variation, a comprehensive biomechanical evaluation for the patellar thickness change under intraoperative/non-load bearing condition is still lacking. The current study utilized both computational and experimental testing platforms involving a cruciate retaining TKR system, to approximate the desired intraoperative physiological loading scenario at multiple patellar component thickness levels. In this regard, previous researchers have also speculated that intraoperative knee flexion motion could serve as a predictor of postoperative knee flexion [141-146]. Similar variation tendency with postoperative observations as the patellar thickness changes could also be expected for other biomechanical parameters, including quadriceps tendon force, patellar kinematics and PF contact pressure. From the perspective of clinical application, understanding the intraoperative influence of patellar thickness change would further provide technical reference and guidance, which refer to potential postoperative outcome, for real-time surgical measuring instruments. In addition, as a prevalent clinical solution for patellofemoral maltracking [147], lateral retinacular release (LRR) procedure was also introduced in the current study as variation to the standard case for more clinically-relevant consideration.
CHAPTER 5  COMPUTATIONAL INVESTIGATION ON EFFECT OF PATELLAR COMPONENT THICKNESS ON QUADRICEPS TENDON FORCE AND PATELLOFEMORAL CONTACT PRESSURE

This chapter focuses on the effect of patellar component thickness on quadriceps tendon force and patellofemoral contact pressure using computational finite element modeling method. Typical FE model consists of three major parts: the geometric modeling, material characterization and definition of boundary conditions.

5.1 Methods

5.1.1 Establishment of the Geometrical Model

In this project, a natural knee joint was first modeled including bones, muscles, ligaments and tendons. The 3D outline of the knee joint articular surface was kept as positioning reference for the procedure of bone resection and TKR implant insertion.

To realistically represent the geometrical components, the current study utilized imaging reconstruction toll (Mimics, Materialise NV., Leuven, Belgium) by constructing and converting a series of 2D axial MRI images into a 3D CAD geometry. The MRI images, obtained from publicly available database [148], were taken from a specific cadaver specimen (Female, Caucasian, age 81, weight 63 kg, height 1.67 m). For 3D reconstruction, anatomic features of thighs were outlined and segmented at each slide,
and then all the slides were stacked in sequence in the 3D space to generate CAD geometry.

Based on previous studies, several common modeling strategies were applied for the current FE model. Geometry of ligaments was simplified based on the mechanical functions and material properties. During the reconstruction process, isolated 3D ligamentous structures cannot be accessed due to the difficulty in segmentation, therefore, geometrical approximations, such as 1D connectors [149-152] or springs [153] and 2D membranes with regular shape were applied instead in previous simulation projects [154, 155]. Also, additional surface smoothen was required to remove individual’s abnormal physical features, e.g. osteophytes and cysts. Tissue microstructures were not involved in the knee joint model to avoid extreme dimensional variance among modeling components. Depending on specific focuses, such microstructures can be represented by compromised material characterization for equivalent material mechanical behavior.

![Figure 5.1 Axial MRI scan of human thighs obtained from public database [148]](image)

Previous knee joint modeling cases majorly selected force as inputs to predict tibiofemoral and patellofemoral motion. Active muscular restraints, including quadriceps and hamstring forces, were represented by force vectors at attachment sites [149-152], so specific muscle geometry and structural characteristics were neglected. For components
providing passive constraints, e.g. ligaments and tendons, simplified geometries were chosen for 3D representation. According to several previous studies focused on patella, the overall tibiofemoral motion was calculated by applying equivalent forces/moments obtained from experimental acquisition on femur and tibia, thus soft tissue kinematic constraints can be in general eliminated [138, 154, 155]. The current FE model was displacement-controlled, which is similar with the study conducted by Halloran et al. [111], but more straightforward and accurately-controlled. The presented geometrical model only consisted of major kinematic constraints specifically for patellofemoral joint.

The quadriceps muscles and quadriceps tendon have been numerically identified as significant and consistent contributor to patellar stabilization [156], so simplified geometrical modeling was avoided. In addition, due to the proximal attachment site of rectus femoris, which is the anterior inferior iliac spine located right above the hip joint, the pelvis bone was also involved in the geometrical model with femur, tibia and patella.

Even though the medial patellofemoral ligament (MPFL) is relatively thin, it has been reported as the primary passive constraint to patellar lateral motion, especially when the knee is close to extended (less than 45° flexion) to prevent lateral subluxation [31, 38, 157, 158]. Also, within the lateral patellar compartment, iliotibial tract functions to resist a medi ally directed force to the patella [18, 27, 40], in addition to the lateral retinaculum which runs along the lateral border of the patella and represents confluence of many structures [25]. Hence, the retinacular structures were mechanically required for the geometrical model.
Based on some previous FE models, the medial and lateral trilaminar soft tissue compartments were usually simplified. MPFL and the lateral patellofemoral ligament, a symmetrical retinacular structure balancing the MPFL, were modeled in general to represent retinacular kinematic restraints [138, 149, 150, 154, 155]. In the current study, to model, reinforced retinacula were also introduced besides MPFL and LPFL to model the capsular restraining along the proximal-distal direction. Consequently, to maximally keep the normal soft tissue constraints with appropriate simplification, anatomical components selected in the presented geometrical model included the quadriceps muscles, quadriceps tendon, media and lateral patellofemoral ligaments and retinacula.

A commercially available TKR system (Left knee, Triathlon®, Stryker Orthopaedics, Mahwah, NJ) was incorporated into the natural knee joint model. The insertion procedure followed specific surgical guideline. Also, knee joint alignment was adjusted following surgical protocols discussed in Chapter 3: 5-7° angle between the femoral anatomic axis and connecting line of the knee, hip and ankle joint centers within the coronal plane (to achieve flat linear profile of the tibial plateau), 2° femoral external rotation and 3° tibial posterior slope were involved for bone resection, to release the over-
tensed soft tissue structures at the medial knee compartment and facilitate deep flexion knee joint motion [91].

CAD geometry of the TKR device was obtained through 3D laser-scanning. To closely replicate the outline features, edge smoothen was processed to remove inconsistent geometries due to scanner error. The insertion procedure of the Triathlon TKR device was conducted using pre-processing software HyperMesh (v11.0, Altair Engineering Inc., Troy, MI), which was also employed for geometrical cleaning and meshing. The surgical procedure followed the same resection reference in the modeling environment, and planar surface cutt was conducted to obtain the desired bony resection. Furthermore, considering the geometrical irregularity of biological components, 3D solid tetrahedron elements which have excellent geometrical flexibility were recruited to computationally discretize the complex physical domain (Figure 5.3).

![3D representation of the FE geometrical model](image)

**Figure 5.3 3D representation of the FE geometrical model**

5.1.2. Material Characterization

For the FE simulation, several hypotheses were given for material modeling. though it is believed that anisotropy and heterogeneity could better define the real
material properties of biological components, most previous modeling-based TKR studies assumed material symmetry and isotropy for bones and soft tissue structures to approximate the overall mechanical material behavior for simplicity. Specifically, for bone material modeling, the isotropic linear elastic material property has been widely implemented [149-155, 157]. In the current study, the patella bone was also modeled as isotropic linear elastic material, and quantified by material parameters of density, Young’s modulus and Poisson’s ratio [158, 159]. The femur and tibia, of which the motion was defined by directly prescribing time-dependent displacement, were modeled as undeformable rigid bodies with respect to the central points of the femoral condyle and tibial plateau, respectively.

The TKR system consists of 4 components, the femoral, tibial components, tibial insert and patellar component (button), of which the femoral and tibial components were made of CoCr and Titanium alloys, respectively, and the tibial insert and patellar button were made of crosslinked UHMWPE (X3®, Stryker Orthopaedics, Mahwah, NJ) [160]. Material data for ASTM F75 and F136 was used for the metallic material behaviors of CoCr and Titanium alloys (Table 3.1).

**UHMWPE Material Modeling**

As mentioned above, the specific crosslinked UHMWPE was produced by compression-molding GUR1020 UHMWPE resin, and undergoing three consecutive gamma irradiation cycles (dosage: 30 kGy) followed by annealing at 130 °C for 8 hours [160]. Compared with untreated UHMWPE, as McKellop et al. noticed, gamma irradiation process could cause slight increase of the material’s yield strength, but
moderate decrease of ultimate strength and significant reduction of toughness/failure elongation [161]. From modeling standpoint, even though several TKR studies still utilized data of conventional UHMWPE for modeling the tibial insert, it is critical to recognize the difference between material properties of crosslinked and uncrosslinked UHMWPE. Typically, UHMWPE can be modeled as non-linear elasto-plastic material [110, 111], with rate-dependent viscoelastic behavior due to coexistence of crystal and amorphous phases at body temperature.

According to previous FE studies for TKR tibial components, UHMWPE presented similar $\sigma$-$\varepsilon$ curves under tensile and compressive loading: elastic material deformation is linear and nearly incompressible with Poisson’s ratio of 0.46, and finite rate-dependent plastic deformation occurs afterwards till total strain of 0.12 [109, 162].

For monotonic loading at single constant strain rate, the classical rate-independent deviatoric plasticity theory with a von-Mises yield criterion (Equation 5.1, where $\sigma_{ij}$ are components of Cauchy stress tensor) followed by isotropic hardening can adequately describe the actual polymeric material behavior [112, 162, 163]. Besides, the featured rate-dependent UHMWPE plastic behavior can be specified by elasto-plastic true stress-strain testing data at multiple strain rates, by which the real time material behavior can be calculated by linear interpolation based on given curves. Hence, for the current model, multiple UHMWPE $\sigma$-$\varepsilon$ data at different strain rates from experimental uniaxial testing was incorporated.

$$\sigma_v = \sqrt{\frac{1}{2}[(\sigma_{11} - \sigma_{22})^2 + (\sigma_{22} - \sigma_{33})^2 + (\sigma_{33} - \sigma_{11})^2 + 6(\sigma_{12}^2 + \sigma_{23}^2 + \sigma_{31}^2)]} \tag{5.1}$$
Bergstrom et al. and Sobieraj et al. developed UHMWPE processing protocols using GUR 1050 to approximate X3® production process [164, 165] and conducted uniaxial tensile-to-failure tests within simulated human body physical environment at two loading rates (30 and 150 mm/min), which provided verified material testing data for X3® UHMWPE modeling in the current study.

Due to the biphasic material property, the viscoelastic behavior of UHMWPE could also affect the patellofemoral joint kinematics and PF joint contact pressure. It has been noted that the material resistance to creep, which is typical viscoelastic behavior, is altered after the gamma irradiation crosslinking process [166, 167]. In the current FE model, more experimentally-accessible stress relaxation testing data was employed to simulate viscoelastic behavior of UHMWPE. Khan et al. obtained uniaxial compressive testing data for a similar crosslinked UHMWPE at -5% strain and 3 predefined strain rates in room temperature, which provided time-dependent variation of normal stress, and was employed to estimate elastic modulus variation of X3® for the presented FE model [168].

Furthermore, time-dependent shear modulus variation was required by the modeling system for viscoelasticity definition, which can be converted from the aforementioned uniaxial testing data based on Equation 5.2. $G_R$, $E_R$ and $\nu$ denote the real-time relaxation shear modulus, Young’s modulus and Poisson’s ratio, respectively. The constitutive equation for material viscoelastic modeling, Equation 5.3, was further approximated utilizing Prony series expansion (Equation 5.4), and time-based $g_R$, which
is normalized shear relaxation modulus quantitatively defined as $G_R/G_0$ ($G_0$ refers to initial shear modulus) was ultimately utilized as model inputs [114].

$$G_R(t) = \frac{E_R(t)}{1-2\nu}$$

$$\tau(t) = \int_0^t G_R(t-s)\dot{\gamma}(s)ds \quad (5.2 \text{ & 5.3})$$

$$g_R(t) = 1 - \sum_{i=1}^{N}\bar{g}_i^p(1-e^{-\frac{t}{\tau_i^G}}) \quad (5.4)$$

**Muscle/Tendon/Ligaments Material Modeling**

Under intraoperative condition, passive stretch is the dominant material behavior for muscles, and according to experimental observation, material stiffness developed in muscles is a changing ratio to the internal tension due to passive stretch [169, 170]. Similarly, non-linear material behavior was also noticed for tendons and ligaments [171, 172, 173], of which the material stiffness highly depends on relative orientation and movement of collagen fibrils. Therefore, by summarizing previous modeling studies, isotropic hyperelastic models were selected as the most accurate constitutive option to describe the passive non-linear/large deformation behavior of muscles, tendons and ligaments [149-152, 154, 155]. For numerical implementation, uniaxial material testing data from previous research attempts were used to describe the nominal stress-strain relation for determination of detailed constitutive model.

Compared with younger patients, due to the progressive deterioration of collagen fibers and replacement by elastin content along with ageing, tendons in older adults are more compliant, slack and less stiff [174]. Given the structural and functional similarity between tendons and ligaments, same alteration could occur on the ligamentous
mechanical property for elderly adults. Muscle strength and stiffness could also have significant reduction because of weakening of muscular architecture induced microscopically by loss of sarcomeres in parallel and series (sarcopenia) [175], causing macroscopically 22-25% less in quadriceps muscle cross-sectional area [176]. Since most osteoarthritis patients and TKR recipients are in aged population, in the current FE model, material property change due to ageing and pathological influence was also considered.

In the current study, uniaxial tensile testing data from Gras et al. was incorporated into the modeling system [177] to quantitatively specify the strain energy density function and material-specific parameters for hyperelasticity definition of muscles. Specific multiplier was then applied to the testing data for simulation inputs based on aforementioned experimental observation for muscle’s ageing effect. Utilizing in vivo ultrasonography for 18 elderly subjects (10 F, 8 M), Reeves et al. investigated the mechanical property of quadriceps tendon and explicitly involved consideration regarding the ageing effect [174], while Bonifasi-Lista et al. obtained uniaxial and shear material stress-strain curves for ligament using 10 specimens from 5 cadavers (3 M, 2 F) with consistent senile age bracket [173]. These two studies provided applicable reference material data for modeling the quadriceps tendon, patellofemoral ligaments and retinacular structures in the current FE model.

In addition, multipliers signifying the effect of osteoarthritis were also needed for compromising modeling inputs. Detailed numerical effect osteoarthritis on tendons and ligaments remains undetermined. However, osteoarthritis causes disuse muscular atrophy and microstructure damage, and previous study has specified strength decrease by 9.49%
(36.9 to 33.4 lb/ft) for female, and 2.63% (53.2 to 51.8 lb/ft) for male. Therefore, such
influence was also involved as multiplying factor for muscle passive property alteration
[178, 179].

Furthermore, because of the intracellular and extracellular existence of water,
muscle, tendon and ligaments can react viscoelastically to passive stretch [172, 173, 180],
presenting time/rate-dependent behavior. Material-specific data from experimental stress-
relaxation tests were gathered and incorporated for describing the time/rate-dependent
viscoelastic behavior [173, 181] via the same approach introduced in the previous section
for numerical implementation.

5.1.3 Acquisition of Knee Joint Kinematics and Definition of Boundary Conditions

To define dynamic displacement boundary conditions for the FE model,
kinematic acquisition testing was conducted using the force-controlled Instron-Stanmore
knee simulator (Instron Corp., Canton, MA) for obtaining the 0-60º knee
flexion/extension kinematics during a gait cycle following ISO standard 14243-1.
Detailed experimental setup and testing specifications are illustrated in Chapter 3, Section
3.2.1.

During the simulator testing, two high-speed video cameras (Phantom V 5.1,
Vision Research, Inc., Wayne, NJ) were perpendicularly placed to obtain the sagittal and
transverse views for the femoral and tibial kinematics during knee flexion/extension.
Reflective markers were attached to femoral and tibial assemblies, and the time-
dependent positions of markers, as well as identifiable reference points (e.g. rotation
center of the femoral arm and middle point of the tibial cup edge) were collected during 5
complete walking cycles and recorded for calculating the rigid body motions of both femoral and tibial components. Motion views were captured at a resolution of 1024x1024 pixels and a frame rate of 250 fps.

The motion tracking video (sagittal/transverse view) was further discretized into a series of images in time sequence, by which individual frames were extracted every 0.004s and digitally read through by a customized program. 2D pixel-based coordinates of the desired marker points within each image at various time frames were recorded and then scaled back to the realistic metric dimension.

For the femoral component, one marked point was selected in addition to the flexion center of the femoral bracket (Figure 5.4a). The flexion was quantified as angular motion of the rigid linear link connecting the two points relative to the standard anterior-posterior direction. Moreover, for the tibial component, in addition to the middle point of the tibial cup edge, another marker was located at the edge corner of the cup holder, which was not blocked from the camera view by femoral component during flexion motion (Figure 5.4b).
Captured coordinates were then transferred to the FE model. Position of the femoral flexion center relative to the TKR femoral component at medial-lateral, anterior-posterior and proximal-distal directions was numerically determined in the FE model based on original design of the fixation phenolic block. According to the relative distance calculated from imaging process, the other femoral marker was also located in the FE model. Initial distances between the tibial cup center and the two marker points were calculated based on cup dimensions and imaging processing. By firstly locating the cup center at the geometrical center of tibial plateau along medial-lateral and anterior-posterior directions, the marker points were also placed in the FE model with original position relationship maintained.

Within the 3D frame, medial-lateral, anterior-posterior and proximal-distal directions were aligned with standard X, Y, Z axes. As Figure 5.5 shows, Point 1 had the same X coordinate with the femoral flexion center, while Point 2 and 3 had the same Z
coordinates with the tibial center. As Point 1 got repositioned during knee flexion, the changing angle between the standard Y direction and the connecting vector from the spatially-fixed flexion center to Point 1 was defined as rotational displacement of the femur. For tibial motion, which included internal-external rotation and anterior-posterior translation, independent rotational and translational displacement boundary conditions were assigned to the geometrical center, and its position relative to Point 2 and 3 throughout motion cycle were maintained constant, so real-time position of the tibial geometrical center was determined through triangular relationship among the 3 points (Figure 5.5), since real-time positions of Point 2 and 3 were known from the imaging processing. The internal-external rotational displacement was specified using the angular change between directions of the standard Y axis and the connecting vector between tibial rotation center and Point 2 or 3, and the anterior-posterior translational displacement was specified as time-dependent Y coordinate change of the tibial rotation center. The kinematic boundary condition was accordingly defined in the FE model using time-based amplitudes of the angular and translational displacement.
Figure 5.5 Marker points from the Instron-Stanmore simulator were placed in the FE model for specification of femoral and tibial rigid body motions.

Quasi-static finite element analysis was conducted using Abaqus/Explicit (v6.11, Simulia Corp., Providence, RI). Penalty contact algorithm was applied in the general modeling domain to prevent inter-penetration and interference among geometrical components. To calculate the quadriceps tendon force, from the standard FE outputs, the internal nodal force vectors within the quadriceps tendon structure were summed together at X, Y and Z directions, and the resultant internal force magnitude was equivalent with the desired quadriceps tendon force. Time-dependent peak patellofemoral contact pressure and PF contact pressure distribution were also exported.

Moreover, using the geometrical morphing function in HyperMesh, four additional patellar components with various thickness levels (±1, ±2 mm in addition to the neutral case) were created (Figure 5.6), thus generating 5 FE models in series for further evaluation.

5.2 Results
During the simulation, corresponding to the intraoperative loading condition, displacement boundary conditions were only modeling inputs, without consideration of body weight. Typical gait cycle involves 0°-60° knee joint flexion, and therefore, quadriceps tendon force, patellofemoral contact pressure and other biomechanical parameters obtained at each flexion angle were exported from the FE model. Variation of the quadriceps tendon force, as well as peak patellofemoral contact pressure across entire contact area was presented in Figure 5.7a and b, respectively.

![Morphed patellar component models at different thickness levels: (a) -1 mm, (b) neutral and (c) +2 mm](image)

Von-Mises stress distribution was obtained throughout the volume of the quadriceps tendon during continuous 0-60° knee flexion. In all 5 cases, the peak stress occurred at 60° flexion, and was observed at the anterior attachment site between the quadriceps tendon and patella. As Figure 5.8 demonstrates, for the case with original patellar component thickness, the maximum von-Mises stress was 5.85 MPa.

All five cases presented similar monotonic curve patterns for variation of quadriceps tendon force during knee joint flexion. All cases reached the peak at 60° knee flexion. Generally, it was noted that as patellar component thickness increases, reduction of the total quadriceps tendon force can be well expected within the flexion range between 0° and 60°, whereas opposite conclusion can be drawn as patellar thickness
decreases. For the neutral case, the maximum quadriceps tendon force value was 527.14 N, while this magnitude for other cases were 502.42 N (+1 mm case), 413.81 N (+2 mm case), 657.37 N (-1 mm case) and 723.10 N (-2 mm case).

As Figure 5.7b demonstrates, variation of the peak patellofemoral contact pressure collected from different cases also presented similar curve patterns. Maximum patellofemoral contact pressure value during knee joint motion cycle was found at 60° flexion, whereas an additional local maximum was recorded between 5° and 10° flexion. Fundamentally, lower patellofemoral contact pressure value was seen in thinner patella cases, while as patellar thickness increases, the contact pressure magnitude gets amplified. For the neutral case, peak patellofemoral contact pressure was 5.67 MPa, and the corresponding value obtained from other cases were 6.24 MPa (+1 mm case), 6.35 MPa (+2 mm case), 5.09 MPa (-1 mm case) and 4.21 MPa (-2 mm case).

![Figure 5.7 Variation of (a) quadriceps tendon force and (b) peak patellofemoral contact pressure during knee joint flexion from 0°-60°](image-url)
Figure 5.8 (a) Computational representation of the knee joint at 60° flexion and (b) Quadriceps tendon reached the maximum deformation at 60° knee flexion

Figure 5.9 further shows the patellofemoral contact pressure at 5° and 60° knee flexion. Different contour patterns were noticed. In Figure 5.9a, the patellofemoral contacting site was proximal to the geometrical center of the patellar articular surface, while in Figure 5.9b, concentrated contact pressure distribution was found at the center of the patellar domed surface.

5.3 Discussion and Conclusion

The current study utilized computational finite element modeling to numerically evaluate the effect of patellar component thickness on the quadriceps tendon force and
patellofemoral contact pressure during knee joint flexion from 0° to 60°. According to simulation results, a thicker patella after TKR surgery tends to be beneficial in reducing the quadriceps tendon force, which is reasonable since the moment arm of the extensor mechanism increase, therefore, less force would be required to complete knee flexion/extension motion. However, a thicker patella is closely associated with amplified patellofemoral contact pressure. This could attribute to the clinical complication “overstuffing”, and is often associated with tensed medial and lateral retinacula. The resultant force from both medial and lateral retinacular structures gets enlarged with a thicker patella, which compresses the patella against the trochlea articulation. Consequently, higher patellofemoral contact can be anticipated.

Based on the current preliminary computational study, a further experimental study would be desirable for validating the FE simulation results and achieving more comprehensive understanding regarding the biomechanical effect of patellar thickness variation on detailed patellar kinematics and patellofemoral joint function during a wider range of knee flexion.
CHAPTER 6 DEVELOPMENT OF A NOVEL UHMWPE PATELLAR SENSOR FOR MEASURING DYNAMIC PATELLOFEMORAL CONTACT PRESSURE

6.1 Introduction

Instances of anterior knee pain and patellar fracture are more and more widely considered as significant complications following total knee replacement (TKR). Specifically, the patellofemoral joint contact pressure and contact area, associated with the geometrical conformity between the articular interfaces, directly correlate with the polyethylene wear and deformation of the patellar component [182]. Hence, real-time measurement of patellofemoral joint contact condition is significant in assessing clinical outcomes and implant long-term success.

In recent years, of the digital pressure-sensing systems that have been developed over the past decade to address such concerns, Tekscan (Tekscan Inc., South Boston, MA) system has been widely used in biomedical, especially orthopedic, applications [183-186]. In Tekscan’s matrix sensors, rows and columns of conductive ink strips intersect to form sensels™, which detect force applied in a dynamic, real-time pattern according to variance in the sensel’s electrical resistance [187]. Since each sensel is an isolated measuring unit, pressure distribution can be collected from each unit, mapped and represented as contours in the graphical user interface in real-time. Although the Tekscan system’s robust, durable performance has been widely reported [187-189], studies also
have noted a few deficiencies: excessive cost (>\$10,000) [190], relatively large digital drift [188], and the necessity of recalibration to accommodate various applications [189]. Additionally, Tekscan systems introduce a different material of 100 \( \mu \text{m} \) thickness in between the joint-contact surfaces, which can alter the original contact topology [191, 192] and increase the possibility of wrinkle-related sensor damage caused by geometrical mismatch between the curved joint surface and the flat sensing film.

In order to address the need to experimentally measure the dynamic contact pressure distribution on the UHMWPE patellar button without changing the mechanical and tribological properties of the articular surface, a unique, low-cost sensor technology was developed. The sensor’s mechanical properties are identical to those of UHMWPE, and it can provide dynamic measurements of contact-pressure distribution throughout the motion cycle. To validate this sensor technology, an UHMWPE composite with modified electrical properties was processed and integrated to domed patellar geometry to evaluate its measurement precision in comparison with the widely-used Tekscan K-scan\textsuperscript{TM} sensor system within a standard mechanical testing frame.

6.2 Methods

As noted above, the current study utilized a novel pressure-sensor technique, which was prototyped based on the geometry of domed patellar component for performance validation before testing. Compared with the tibial insert, the morphology of the patellar component used in the current study was more accessible from a machining standpoint. However, surface profile of the patellar component is less conformal for joint contact. Thus, to obtain real-time measurement, a sensing unit with greater sensitivity and
performance is required. Therefore, the patellar pressure sensor could serve as a reliable prototype platform to validate the precision and accuracy of the novel sensor technology. A rectangular block (82.55 mm L x 57.15 mm W x 19.05 mm H) was formed by compression-molding GUR 4150 UHMWPE resin powder in a stainless steel mold at 210 °C at a pressure of 10 MPa for 20 minutes and then 44 MPa for 40 minutes [193]. The same processing conditions were used to mold many 1.59 mm diameter pegs of blended 8% carbon black (CB)/UHMWPE composite sensor material [107]. Then 5 columns by 5 rows of 1.59 mm diameter holes on a 2.54 mm grid for the block were machined (Figure 6.1a). After the UHMWPE-composite pegs were inserted in the holes, the blocks were compression-molded again under the same processing conditions to fuse the composite pegs with the virgin UHMWPE blocks. This created a completely fused, solid block of mainly virgin UHMWPE with localized areas of the UHMWPE-composite sensor material. Because the bulk mechanical properties of the composite sensor material are not significantly different from those of the virgin UHMWPE [194], the instrumented blocks could be machined into domed patellar-button geometry following manufacturing specifications. This ensured that the instrumented implant had contact geometries identical to those of the corresponding production UHMWPE component implanted in patients. The selected geometry was the Stryker Triathlon® patellar component (Stryker Orthopaedics, Mahwah, NJ).

After the contacting surface was machined, wires were attached to the unmachined backside of each component. The wires connected each sensing point to its corresponding channel in a customized data acquisition system, which was used to filter
and collect the raw sensor outputs. The system consisted of dual 256 channel customized analog multiplexer arrays and the necessary filtering amplifiers to output a single-channel data stream of voltage values that corresponded to the instantaneous contact pressure measured at each sensing point. This single-channel data stream was clocked into a high-speed A/D converter card on the computer. An additional wire was attached to the femoral component of the implant so that an excitation voltage could be applied. Once the data was in the computer, a graphically intensive software interface (Labview 2010, National Instruments Corporation, Austin, TX) was used to store, visualize, and analyze the dynamic contact pressure and its distribution.

Figure 6.1 (a) 3D view of block with enlarged grid pattern to illustrate how blocks were fabricated; (b) Top view of the machined and instrumented prototype of the UHMWPE patellar sensor

To validate the precision and accuracy of the proposed novel sensor technology, a comparative test was conducted using the UHMWPE patellar sensor and the Tekscan K-scan™ 4000 system for orthopedic application. Based on the desired testing condition and loading limits, both pressure sensors were calibrated by uniaxial compression testing using an Instron hydraulic material-testing machine (Model 8874, Instron Inc., Norwood,
MA). The Tekscan K-scan™ 4000 sensor contains 572 individual sensing elements in a 26 row by 22 column matrix (Figure 6.2a) [195]. As recommended by the product manual, the K-scan™ sensor was calibrated based on an exponential curve fitted with 2 subsequent force inputs at 200 N and 800 N, corresponding to 20% and 80% of the maximum desired loading. To maintain a consistent calibration curve for every sensing element, the Tekscan sensor was placed underneath a deformable rubber bladder to apply evenly distributed compressive loading on the sensel matrix.

A series of loading points was also achieved for the novel UHMWPE patellar sensor to fit the calibration curve. A geometrically conformal stainless-steel pressing indenter with known internal contacting area (Figure 6.2b) was designed to ensure equally distributed pressure along the sensor’s curved surface. Compressive loading was linearly and continuously increased from 0-5000 N, corresponding to equivalent 0-15 MPa of the applied pressure. Pressure magnitudes calculated based on Instron’s load-cell readings collected every 500 N, were numerically correlated to real-time voltage responses at each sensing point to define the calibration function.

Once calibration was completed, the comparative-sensor was validated. The Tekscan sensor was placed on the top of the curved patellar surface, aligned, and secured within a customized PE fixture in the middle of the Instron testing frame (Figure 6.2c). A customized indenter was molded to comply with the domed-surface profile and then implemented, which has a circular cross-section that fully covers the top of the 5x5 CB/UHMWPE square-shaped sensing matrix. Uniaxial compression testing was then performed with a dynamic loading ramp range from 0 to 900 N. Real-time peak pressure
values and the corresponding contact-pressure distribution were collected and exported from GUIs of both sensors approximately every 100 N.

Figure 6.2 (a) The Tekscan K-scan™ system used in current study; (b) Customized stainless-steel indenter utilized in patellar sensor calibration; (c) Experimental setting for comparative sensor validation testing

The testing procedure was repeated 4 times for statistical significance. A generalized linear regression analysis with maximum likelihood estimation was carried out in SAS (v9.0, SAS Institute Inc., Cary, NC), which fits statistical models to data with nonconstant interaction and random response for variance evaluation [196]. Because inconsistency existed among all trials regarding the collected loading levels during the dynamic testing process, the proposed statistical model included fixed variables of the compressive-load readings acquired by Instron’s load cell and 2 sensor outputs, plus random variables from different trials and potential interaction between the sensor
performance and trials. The statistical equivalence between pressure measurements of the novel patellar sensor and the Tekscan K-scan™ system was further assessed.

6.3 Results

The variation in peak contact-pressure magnitude during the dynamic loading ramp from all 4 trials, the Tekscan K-scan™ system, and the UHMWPE patellar sensor made from the novel sensor technology demonstrated close curve patterns (Figure 6.3). Considering the Tekscan sensor readings as the baseline measurements, in Trial 1 the maximum difference in pressure measurement between both sensors was 1.82 MPa (around the compressive loading of 600 N), while for other trials, this value was found smaller (0.89, 0.62 and 0.90 MPa), around the loading of 800 N, 500 N and 600 N, respectively. A minor change in the relative alignment between two sensors was observed after the initial testing trial, leading to a slight measurement discrepancy between Trial 1 and Trial 2. In general, according to the generalized linear regression analysis with maximum likelihood estimation procedure based on all 4 trials, the calculated statistical difference between the peak pressure-variation curves obtained by both sensors was insignificant; this was further supported by a p-value of 0.8955.

The pressure distribution presented by both sensors have the same contour patterns: At all loading levels, peak value was noted at the central position, and contact pressure decreased in gradient at peripheral positions, which is consistent with the theoretical expectation (Figure 6.4). Contact area displayed in the Tekscan system is equal to the indenter’s cross-sectional area, which is slightly bigger than the sensing area (in the current case, the contact area since all sensing points were in contact) of the
UHMWPE patellar sensor. To compensate for the potential insufficiency of the sensor resolution, linear interpolation was imposed among CB/UHMWPE sensing points for continuous presentation of pressure distribution.

Figure 6.3 Peak pressure-variation curves obtained by both Tekscan K-scan™ system and UHMWPE patellar sensor

6.4 Discussion and Conclusion

The sensor technology presented in this study is based on a composite of UHMWPE with modified electrical properties that allow it to quantify the contact pressure being applied to its surface. An excitation voltage is applied to the metallic counterface, and upon contact between the surface and each sensing point, an electrical current proportional to the contact pressure will flow through the sensing point. The electronics measure this current at a high rate at each sensing point, which gives the contact-pressure distribution on the implant surface. By continually measuring these
signals, the dynamic contact-pressure distribution can be measured over the duration of the load cycle. Because the exact location of each sensing point on the surface of the instrumented implant is known and because the geometry of the grid pattern of sensing points dictates a given area covered by each individual point, the dynamic contact area and its exact location on the surface can be determined by monitoring the sensing points that show contact. The spatial accuracy of the sensor is therefore determined by the size of each sensing point and can be tailored to the specific application.

Figure 6.4 Pressure distribution at 3 real-time loading levels presented by Tekscan (left) and UHMWPE patellar (right) sensors
As in comparative testing conducted by Bachus et al. [187], the sensor-prototype validation experiment in the current study utilized standard uniaxial compressive testing for simplicity. Through the comparison with the Tekscan K-scan™ system, the patellar sensor demonstrated close performance in accurately and quantitatively capturing the pressure change and the real-time pressure distribution during a ramp-based dynamic loading process (Figure 6.3, Figure 6.4). A minor measurement discrepancy in pressure-variation pattern was noted between Trial 1 and Trial 2, which could be attributed to the alignment adjustment of the stacked two sensors via elastic deformation in response to the initial loading. For the rest trials, measurement repeatability was well maintained by both sensors. In summary, based on the comparative sensor-prototype validation testing and the technical versatility of the generic sensing mechanism, the accuracy, precision, and measurement reliability of the novel sensor technology in the current study can be guaranteed in the application of tibiofemoral contact evaluation.
CHAPTER 7  EFFECT OF PATELLAR COMPONENT THICKNESS ON PATELLAR KINEMATICS AND PATELLOFEMORAL JOINT MECHANICS FOLLOWING TOTAL KNEE REPLACEMENT---A CADAVER STUDY

This chapter focuses on the effect of patellar component thickness on patellar kinematics and patellofemoral joint mechanics using cadaver study.

7.1 Methods

A patient-specific cadaver specimen (Left leg, Male, Caucasian, age 49, weight 51.25 kg, height 1.75 m, KL score for symptomatic OA 2) was utilized in the current study for in-vitro simulation. The specimen was collected postmortem, stored at -20°C until one day prior to dissection. Skin, fascia and majority of the muscle belly present on the leg were removed, leaving only the tendinous structures for quadriceps muscles (vastus medialis (VM), vastus intermedius (VI), vastus lateralis (VL) and rectus femoris (RF)) and hamstring muscles (semmembranosus (SM) and biceps femoris (BF)) (Figure 7.1a). The integrity of the quadriceps tendon and knee joint capsule, as well as enclosed ligamentous structures (collateral ligaments, cruciate ligaments and retinacula) were maintained. During the dissection process, the specimen was covered with absorbent pads soaked in 0.1M solution of saline to maintain moist state [197]. Upon completion of dissection, the cadaver leg was stored at -20°C once again until the day of testing and refrigerated overnight before testing (slightly above 0°C for 12 hours) to thaw it out.
slowly [197]. On the testing day, the cadaver was soaked in 0.1 M saline solution for approximately 60 minutes, to increase muscle hydration and to enhance the tissue’s hydration-dependent response [197].

In the current study, a custom knee testing rig was implemented to investigate the kinematics of the TKR-inserted cadaveric knee joint (tibiofemoral and patellofemoral joints) during knee flexion/extension simulation. It equivalently initiates the knee flexion/extension process by imposing displacement to the quadriceps muscle-tendon units, with a balancing hamstring force applied which is consistent with the co-contraction condition in natural human knee joint. The Clemson Knee Rig also models hip and ankle assemblies, and by exerting predefined position and orientation for the hip and ankle joints, it enables six degrees of freedom for a realistic knee joint motion [198].
The specimen’s bone shaft was trimmed 120.65 mm off proximally from the ankle joint and 130.18 mm distally from the hip joint for installation onto the rig. Epoxy resin was used for potting both ends of the specimen in the center of the cup-shaped rig fixtures. Alignment of the cadaver specimen was further conducted utilizing radiographic inspection to achieve flat linear profile of tibial plateau, 5-7° angle between the femoral anatomic axis and connecting line of the knee, hip and ankle joint centers within the coronal plane, and superimposed medial and lateral femoral condylar profiles within the sagittal plane. Preexisting tibiofemoral varus/valgus rotation was prevented in the current case.

A commercial TKR system (Left knee, Triathlon®, Stryker Orthopaedics, Mahwah, NJ) was inserted into the cadaver specimen following the specific surgical guideline (Figure 7.2b). Based on the alignment/resection guideline, 2° external femoral rotation and 3° tibial posterior slope were involved for implant insertion, to release the over-tensed soft tissue structures at the medial knee compartment, and facilitate deep flexion knee joint motion [91]. Compared with the intact patellar thickness, 1 mm extra bony resection was set to obtain an experimental case with reduced patellar thickness. Instead of using the original patellar component with bone cement, a customized UHMWPE patellar button with a matrix of pressure sensing units was used to measure the real-time PF contact pressure (Figure 7.3a) without altering the mechanical and tribological property of the contacting interface. The sensor’s precision and accuracy have been well validated comparing with widely-applied commercial sensors (see Chapter 6) [199], therefore the gauging reliability for PF contact pressure in the current
case can be guaranteed. Specific fixture with screw connections to the bone was designed to mechanically fit the pressure sensor to patella (Figure 7.3b-c).

Figure 7.2 Experimental setup for the patient-specific cadaver specimen: (a) Polaris motion tracking tools were utilized for capturing rigid body motion of 1- femur, 2- tibia and 3- patella; (b) the commercial TKR device was inserted into the specimen; (c) tendinous structures of quadriceps muscles, VM, VI/RF and VL were arranged as prescribed by Sakai et al. [200].

The RF/VI, VM and VL tendons, as well as SM and BF tendons, were adjusted in length and attached to customized tissue clamps (Figure 7.1b), which have zigzag-interface to increase friction along the tissue-clamp contact interface and prevent relative slipping. According to previous studies, during normal knee joint flexion/extension motion, the applied force directions of the 3 major quadriceps muscle bundles are expected to be divergent [200]. By following in-vitro measurement conducted by Sakai et al., the VI/RF tendons were arranged 8.5° medial to the femoral anatomic axis in the coronal plane and 4° anterior to the femoral axis in the sagittal plane; VM was placed
42.50° medially to the femoral axis in the coronal plane and parallel to the femoral axis in
the sagittal plane; and VL was 13.50° laterally to the femoral axis in the coronal plane
and parallel to the femoral axis in the sagittal plane (Figure 7.2c) [200]. All tendinous
structures of quadriceps muscles were ultimately merged together and connected to a load
cell as well as DC motor in series via steel cables and pulleys. Besides, force magnitudes
required to complete knee flexion/extension motion tend to be different among
quadriceps muscles.

Figure 7.3 (a) The customized UHMWPE patellar sensor employed in the current study with the
same geometrical shape as patellar component; (b) CAD model of the specific fixture attached to
patella to fit the pressure sensor; (c) The patellar pressure sensor inserted into the cadaver
specimen with PE shims for adjusting thickness levels
Estimation based on muscle physiological cross-sectional area (PCSA), which is the area of cross-section perpendicular to the muscular fiber direction and quantitatively equal to muscle volume divided by fiber length, revealed that relative force ratio of VM: RF/VI: VL is 2:3:2.5 [149, 150, 200]. Therefore, in the current study, length of the connecting steel cables were further adjusted for 3 quadriceps muscle bundles. Within the unit time span, magnitudes of distance traveled by tendon ends at 3 divergent directions (VM, RF/VI and VL) should follow the aforementioned relative force ratio, maintaining the equivalent relative force inputs.

Four sets of motion tracking tools (Polaris optical tracking system, Northern Digital Inc., Waterloo, Ontario, Canada) made of acrylic and assembled with reflective markers were placed at the global reference frame, femur and tibia (along bone shafts), as well as patella, to optically track the rigid body motion (spatial position coordinates and Eulerian angles) of the three bone segments (Figure 7.2a). The patellar tracking tool was screwed onto the anterior patellar surface while maintaining the orientation parallel with the patellar longitudinal direction, and aligned in line with the proximal-distal center of patellar median ridge at both transverse and sagittal planes [201]. Correspondingly, the center of the tracking tool (local coordinates) was identically matching the geometrical center of the dome-shaped patellar button.

To simulate the non-load bearing intraoperative condition, weights were added to counterbalance the body weight. The cadaver leg specimen was lowered to deep flexion position (100°), and then imposed quadriceps displacement driven by the DC motor was gradually increased at constant speed until extending motion started and towards to full
extension. Real-time PF contact pressure, quadriceps tendon force and bone segmental kinematics were measured and recorded during the extension movement by the patellar sensor, load cell and motion tracking system, respectively. Regarding the SM and BF tendons, equally small amount of loading (10 N) was placed to tighten the structures, and provide counter balance force during knee extension. To preserve effective fit between the patellar button peg and the resected holes for patellar stability, up to 3 mm extra “stuffing” was allowed in between the patella bone and implant component to vary the composite thickness. Meanwhile, minimum residual bony thickness was also maintained. Consequently, the knee joint motion testing was conducted at 3 different patellar component thickness levels: -1 mm, neutral and +3 mm, and replicated three times within each level for average. Since all measurements were based on the same testing frame/specimen, multivariate analysis of variance (MANOVA) was conducted to assess the statistical difference among cases using SAS (v9.0, SAS Institute Inc., Cary, NC). Customized 1-mm thick polyethylene shims with consistent geometry with the sensor fixture were used to vary the patellar component thickness with increment of 1 mm.

Furthermore, in the current study, influence of the lateral retinacular release (LRR) procedure was also involved in evaluating patellar kinematics/kinetics for more comprehensive understanding. According to the procedure provided by Merican et al. [40] and considering the integrity of the specimen after TKR insertion, the lateral retinaculum was incised along the proximal-distal direction, from 20 mm proximal to the proximal end of the patella to the distal pole of the patella. Thick retinacular connection from the iliotibial band to the VL tendon and lateral patellar end was also disrupted [201,
Biomechanical assessment at 3 patellar thickness levels was also conducted for the specimen after LRR procedure performed.

7.2 Results

Generally, natural patellar motion during knee joint flexion/extension refers to 6 degrees of freedom: medial-lateral (M/L) shift, superior-inferior (S/I) translation, anterior-posterior (A/P) translation, medial-lateral rotation, medial-lateral tilt, and flexion [203], among which four are perceived as clinically important motion parameters: the M/L shift, M/L rotation, M/L tilt and flexion [201, 204]. Patellar motion can be physiologically prescribed based on mixed patellar and femoral body fixed axes: M/L shift is essentially patellar movement along the femoral medial-lateral direction relative to the center of the trochlear groove; M/L rotation and tilt are perceived as rotation about patellar anterior-posterior and long (proximal-distal) axes, respectively; patellar flexion refers to rotation about the femoral medial-lateral direction [204] (Figure 7.4).

The Polaris motion tracking system recorded rigid body kinematics, neglecting detailed morphological features, where femur, tibia and patella are represented by reference points corresponding to individual reflective maker tools. Following the predefined coordinate system for motion tracking, M/L shift can be numerically converted to real-time point-to-surface distance from the patellar reference point to an auxiliary datum plane which bisects the trochlear groove and passes through its geometrical center. M/L rotation, M/L tilt and patellar flexion can be approximately transferred as rigid body rotation with respect to y, x and z axes, respectively. Segmental positional and rotational information of the knee at full extension state was recruited as zero reference for real-
time measurement. Due to patellar flexion motion, of which the rotating axis is parallel with the global z axis, time-dependent orientation of patellar long axis was not aligned with the x axial direction, thus captured patellar M/L rotation and tilt angles were estimated from projections at standard XOZ and YOZ planes (Figure 7.4), respectively. However, the recorded real-time Eulerian angle of patella along z axis (instead of patellar flexion angle with respect to femoral medial-lateral axis) showed maximum of 10° in magnitude, and for M/L rotation and tilt, the calculated difference between actual and projected angles was directly associated with cosine of such Eulerian angle, indicating the maximum value deviation ranged up to 1.5%, and therefore, could be neglected.

![Figure 7.4 Patellar motion identified based upon combined femoral and patellar body fixed axes](image)

Detailed patellar kinematics at three different patellar component thickness levels within the normal and LRR cases is presented in Figures 5 and 6. As shown in Figure 7.5, relatively consistent curve trend can be found for patellar M/L tilt, shift and patellar flexion at -1, neutral and +3 mm thickness levels. Intermediate curve pattern variation
can be observed for patellar M/L rotation (Figure 7.5b). For the case with -1 mm patellar thickness, the lateral tilt angle during the knee joint motion cycle peaked at 20° flexion, and the corresponding value was 8.51±1.45° (Figure 7.5a). The minimum point was found at 90° knee flexion, with value of -1.54± 2.69°, referring to medial patellar tilt. During the entire motion cycle, the patellar tilt (angle) showed monotonic increase between 0 and 20° flexion, decrease between 20 and 90° flexion, and slight pick up around 100° flexion. For the case with original patellar thickness (neutral case), the lateral patellar tilt angle also peaked at 20° flexion, with maximum magnitude of 10.38±0.69°. Overall minimum occurred at 90° flexion as well, showing 1.66±0.27° lateral tilt. Similar curve variation can be noted for the neutral case, which also consisted of 3 phases: monotonic and steep increase (0-20°), gradual decrease with changing slope (20-90°) and pick up (90-100°). Median of the aforementioned two cases were 3.19±1.40° and 5.58±0.39° (at 60° flexion), respectively, while for the case with +3 mm patellar thickness, the median was also found at 60° knee flexion, with the value of 7.14±0.82° Meanwhile, the maximum and minimum magnitudes of lateral tilt for the thicker patella case were 12.48±0.07° at 20° and 2.67±0.94° at 90° knee flexion, respectively. In summary, as patellar thickness changes, patellar tilt magnitude is different (p≈0.02), and the variation curve shifts parallel towards to the lateral direction as patella gets thicker, which simply reveals that patellar (component) thickness increase results in amplified patellar lateral tilt.
Figure 7.5 From the optical motion capturing data, processed intraoperative patellar kinematics during the experimental knee joint movement at 3 patellar component thickness levels, including (a) patellar medial-lateral tilt, (b) patellar medial-lateral rotation, (c) patellar medial-lateral shift and (d) patellar flexion.

For the case with -1 mm patellar thickness, initial medial rotation (negative values) was observed rather than lateral (Figure 7.5b). The medial rotation angle slightly decreased towards to 10° knee flexion, and increased continuously until reaching 100° flexion. The minimum medial rotation angle can be found at 10°, with a value of 0.96±0.67°, while the maximum value was 2.87±1.14°, recorded at 100° flexion. Contrarily, for the case with original patellar thickness, minor curve perturbation was noticed. M/L rotation angle ranged from -0.44±0.49° at 100° to 1.19±0.28° at 20° knee flexion, and majorly lateral patellar rotation was observed during the motion cycle.
Compared with the thinner patella case, the neutral case showed similar curve pattern. During knee joint extension/flexion motion, the patellar M/L rotation was less medialized at lower flexion angles but less lateralized as the angle increased. The neutral case presented median value of 0.77±0.36° at 60° flexion, which is numerically higher than that of the thinner patella case (-1.74±0.77°, also at 60° flexion). For the case with +3 mm patellar thickness, maximum and minimum lateral rotation magnitudes were noted at 20° (1.11±0.09°) and 90° (-0.74±0.02°) of flexion, respectively, while the median value was 0.11±0.19° at 60° flexion, which is lower than the neutral case. The thicker patella case retained similar curve variation pattern with the thinner and neutral cases, yet the local extrem can be more explicitly identified.

As shown in Figure 7.5c, consistent curve trend and solely lateral shift can be perceived for all patellar thickness cases. For the case with reduced patellar thickness, the maximum lateral shift was 3.12±1.18 mm, occurred at 50° knee flexion. Also, within the data group, the median value was 1.99±1.20 mm, found at 70° flexion. From the case with reduced patellar thickness to the neutral case and thicker patella case, the overall variation pattern was well maintained, and corresponding lateral shift magnitudes were amplified. For the neutral case, the global peak lateral shift was 6.42±0.03 mm at 50° flexion, and the median value was 4.95±0.04 mm at 70° flexion. For the case with increased patellar thickness, the global maximum lateral shift was also found at 50° flexion, with the value of 7.54±0.51 mm, while the median value was among the highest in all three cases, 6.37±0.47 mm at 80° flexion. Comparing all three cases, it can be
concluded that increased patellar thickness significantly enhances the patellar lateral shift ($p \approx 0.01$).

As Figure 7.5d shows, monotonic curve variation was observed for the patellar flexion motion. For all three cases, maximum patellar flexion was obtained at the highest flexion angle. The peak patellar flexion angle captured from the cases with reduced and increased patellar thickness, as well as the neutral case were $49.58 \pm 1.47^\circ$, $54.68 \pm 2.02^\circ$ and $52.33 \pm 1.37^\circ$, respectively. In addition, median values of all three cases collected at $50^\circ$ knee flexion, half of the motion cycle, with magnitudes of $18.73 \pm 0.45^\circ$ (reduced patellar thickness), $24.64 \pm 0.35^\circ$ (neutral) and $26.97 \pm 0.46^\circ$ (increased patellar thickness).

Generally, as the patellar component thickness increases, patellar flexion angle varies following steeper ($p \approx 0.04$) and closely linear trend line, reflected by higher real-time slope in the variation curve.

Once the lateral retinacular release procedure was performed, detailed patellar kinematics, including the M/L tilt, rotation, shift and patellar flexion was also captured as reported in Figure 7.6. In general, all curves retained not only the individual curve changing pattern along with the flexion angle, but also the relative curve layout among cases with different patellar thickness levels. Regarding the detailed magnitudes, as depicted in Figure 7.6a, patellar tilt peaked at $20^\circ$ flexion, with the value of $5.68 \pm 0.77^\circ$ for the case with reduced patellar thickness, while negligible medial tilt was found at $90^\circ$ flexion. For the neutral case, the global maximum tilt value was $9.05 \pm 0.69^\circ$. Furthermore, for the case with increased patellar thickness, the overall highest tilt angle was $11.39 \pm 0.75^\circ$. Only lateral tilt was observed for the neutral and thicker patella case.
Compared with Figure 7.5a, Figure 7.6a also reveals that patellar thickness increase is associated with excessive lateral tilt, represented by tendency of the variation curve to parallel translate towards lateral direction. Moreover, at individual thickness levels, evident reduction on lateral tilt angle can be seen in cases with LRR condition than the normal ones. By comparing the median values, significant numerical decrease (%) were 60.8%, 19.8% and 24.8%, for -1 mm, neutral and +3 mm cases. Besides, flexion angles corresponding to major curve turning points were consistent from the normal to the case with LRR performed.

![Graphs showing patellar kinematics](image)

Figure 7.6 From the optical motion capturing data, processed intraoperative patellar kinematics during the experimental knee joint movement at 3 patellar component thickness levels with lateral retinacular release (LRR) procedure performed, including (a) patellar medial-lateral tilt, (b) patellar medial-lateral rotation, (c) patellar medial-lateral shift and (d) patellar flexion.
In comparison with Figure 7.5b, patellar M/L rotation variation curves shown in Figure 7.6b demonstrated more similar pattern, especially for the flexion angle interval from 40° to 100°. For the reduced patellar thickness case, the captured data reported only medial rotation, matching the normal specimen. The maximum medial rotation angle was 1.78±0.56° at 90° flexion, while the median was 1.13±0.72° at 60° flexion. The curve variation can be in general divided into three regions, gradual medial rotation increase from 0° to 40° flexion, closely stationary state from 40° to 80° flexion, and further increase of the medial rotation angle from 80° to 90° flexion. For the neutral case, patellar lateral, rather than the medial rotation, took the predominant role, which was also reflected in Figure 7.5b. The neutral case presented overall peak at 20° flexion (0.57±0.28°), and the thicker patella case had global maximum at 20° as well (0.70±0.02°). Median values were 0.30±0.35° at 30° flexion and -0.25±0.10° at 40° flexion for the neutral and thicker patella cases, respectively. Based on the median curve values, both Figure 7.5b and 6b identified that higher lateral patellar rotation occurred in the neutral case. Also, no significant difference ($p>0.05$) was found for median values between the normal and LRR cases at all three thickness levels.

Slight difference in curve pattern was noticed between Figure 7.6c and Figure 7.5c. After LRR procedure was performed, small amount of medial shift was captured within the neutral and thinner patella cases at lower flexion angles. Similar medial shift was also recognized in the case with reduced patellar thickness at higher flexion angles. The peak patellar lateral shift were 2.23±0.76 mm at 20° knee flexion, 4.17±0.15 mm at 40° flexion and 7.29±0.11 mm at 40° flexion for the case with reduced, neutral and case
with increased patellar thickness, respectively. Within the flexion interval between 20° and 45°, all three cases peaked at the same stage of knee flexion/extension motion. In addition, by averaging all curve magnitudes, mean values of the 3 cases were 0.61 mm, 2.37 mm and 5.74 mm for reduced, neutral and increased patellar thickness cases, respectively. Therefore, consistent with Figure 7.5c, the 3 variation curves also revealed that extensive patellar thickness contributes to extra patellar lateral shift. By comparing the curve averages from both Figure 7.5c and Figure 7.6c, after the LRR procedure, the mean value dropped by 69.3%, 49.1% and 2.8% for -1 mm, 0 and +3 mm cases.

Figure 7.6d reported monotonic increasing trend of patellar flexion angles along with flexion angle increased, which was also seen in Figure 7.5d. For all 3 thickness levels, the maximum patellar flexion angles reached were 48.24°, 50.64±1.37° and 54.05±0.34°, for the thinner, neutral and thicker patella cases, respectively. In terms of curve median values, the aforementioned three cases presented magnitudes in order: 18.77±0.67°, 22.05±0.35° and 26.52±0.13°. This follows the pattern specified depicted in Figure 7.5d, where as patellar thickness levels up, real-time slope of the patellar flexion variation curve increases as well. Likewise, no significant difference (p>>0.05) regarding curve magnitudes was identified before and after the LRR procedure.
Figure 7.7 Variation of quadriceps tendon force during the experimental knee joint motion measured by load cell at different patellar thickness levels for (a) the normal case and (b) case with LRR procedure performed.

Figure 7.7 presents the variation of the quadriceps tendon force measured by load cell, while Figure 7.8 shows the varying maximum PF contact pressure magnitude across the entire contact area. As Figure 7.7 shows, for cases before and after LRR procedure, consistent curve trend was obtained for multiple thickness levels. For Figure 7.7a, the peak quadriceps tendon force values during the entire knee joint motion cycle were 592.08±4.27 N, 575.47±12.13 N and 589.32±8.32 N for reduced, neutral and increased
patellar thickness cases, respectively. Similarly, the peak quadriceps tendon force magnitudes recorded in Figure 7.7b were 578.92±2.27 N, 517.76 N and 520.02±1.85 N for the thinner, neutral and thicker patella cases, respectively. For both sets of curves, the maximum quadriceps tendon force magnitudes were achieved at the highest flexion angle, i.e. 100°. Within Figure 7.7a, compared with the case with reduced patellar thickness, average curve values of the other 2 cases were reduced by 5.75% (neutral) and 8.72% (+3 mm), respectively, whereas the corresponding quantities in Figure 7.7b were 12.45% (neutral) and 15.07% (+3 mm). Statistical analysis based on all curve data has identified the significant difference among cases with various patellar thickness levels within each figure (p≈0.02, normal; p≈0.01, LRR), however, no statistical evidence was available to support a significant difference between mean values of the normal and LRR performed cases (p≈0.4). In general, based on the layout of curves, increased patellar (component) thickness is associated with reduced quadriceps tendon force, which facilitates the knee joint motion. However, local exception should be noticed that at higher flexion angles (>80°), quadriceps tendon force measured in the thicker patella case was in fact higher than for the neutral case.
Figure 7.8 Variation of peak patellofemoral contact pressure value across patellar articular surface during the experimental knee joint motion measured by the customized pressure sensor at different patellar thickness levels for (a) the normal case and (b) case with LRR procedure performed.

The peak PF pressure variation followed quite variable pattern for both normal and LRR cases with maximum pressure values of 1.08±0.14 MPa at 60° flexion, 2.00±0.54 MPa at 50° flexion and 2.33±0.68 MPa at 80° flexion for the reduced, neutral and increased patellar thickness cases, respectively (Figure 7.8a). Similarly, for cases after the LRR procedure, the thinner patella case presented a maximum pressure at 70° flexion with a magnitude of 0.95±0.07 MPa, while highest pressure values occurred at
60° flexion in the other two cases (1.23±0.17 MPa for neutral case and 1.43±0.01 MPa for thicker patella). Median values of the three cases were 0.58±0.11 MPa for the thinner patella, 1.03±0.12 MPa for the neutral case and 1.61±0.26 MPa for the thicker patella for normal case, while the corresponding median pressure magnitudes after LRR were 0.61±0.10 MPa (-1 mm), 0.65±0.17 MPa (neutral) and 0.91±0.06 MPa (+3 mm) (Figure 7.8b). As patellar thickness increases, significantly higher PF contact pressure is generated (p≈0.02). Data have also shown that LRR procedure significantly lowers contact pressures (p≈0.01).

Figure 7.9 presents representative real-time PF contact pressure distribution from cases with various patellar thickness levels. Prior to the LRR procedure, contact pressure contours of the thinner patella case at 60° (trial #1), the neutral case at 50° (trial #2) and the thicker patella case at 80° flexion (trial #3) were displayed. Upon completion of the LRR procedure, contact pressure contours of the thicker (trial #2) and neutral patella cases at 60° knee flexion (trial #1), and the thinner patella case at 70° flexion (trial #3), were demonstrated as well. Figures 9(a)-(f) majorly refer to pressure contours when peak magnitudes were reached within each individual case, and consistent distribution pattern was obtained. The maximum patellofemoral contact pressure was observed at the center of the patellar articular surface dome in all cases, and pressure value decreased in gradient along the radiant direction away from the center position. Higher pressure magnitudes were noticed at peripheral region represented in Figure 7.9c and f. Since these two figures correspond to cases with increased patellar thickness, larger contact
area could be made due to inclined patellar position relative to the trochlear groove, which could be attributed to higher patellar flexion angles during knee joint motion.

Figure 7.9 Real-time patellofemoral contact pressure distribution contours exported from the customized pressure sensor for (a) thinner patella case at 80°, (b) the neutral case at 50° and (c) the thicker patella case at 60°; and after LRR, (d) thinner and (e) neutral patella cases at 60°, and (f) the thicker patella case at 70° flexion

7.3 Discussion

From a clinical perspective, it is important to understand the comprehensive influence of patellar thickness change on patellar kinematics and patellofemoral joint
function, in order to guide selection of appropriate patellar component during the resurfacing procedure. This is especially critical for patients with complicated physiological conditions, for typically it is challenging to maintain the original patellar thickness where either thinner or thicker patellar assembly could likely result. Utilizing patient-specific in-vitro cadaver experiment with customized instruments, the current study provided an extensive overview and full reference regarding the varied patellar component thickness, including specific consideration of patellar M/L tilt, rotation, shift and patellar flexion, quadriceps tendon force and patellofemoral joint contact pressure.

Natural patellar tracking motion depends heavily on the active muscular constraints, passive ligamentous constraints and geometrical congruence. Soft tissue restraints play a significant role in determining the patellar stability before 30° knee flexion when the patella is not fully engaged within the trochlear groove, and effective geometrical constraint is not established. As flexion increases, geometrical congruence primarily controls patellar motion and posture [27]. Analogically, for a TKR-inserted knee joint with a resurfaced patella, kinematics of artificial patellar component tends to also rely on interaction with surrounding soft tissues and geometrical design of the femoral trochlear groove, especially for the intraoperative condition, under which the muscular constraints depend solely on the passive material property.

According to previous study conducted by Amis et al., a summarized pattern of intact patellar tracking starting from extension state includes (a) patellar medial translation till 20° flexion, and gradual lateral shift till 90° flexion; (b) 4°-5° patellar lateral tilt once patella slides into the trochlear groove, and more tilt to around 7.3°
towards to 90° flexion; (c) small amount of patellar M/L rotation (<3°) following a variable trend [205]. Fundamentally, such detailed patellar motion was directly determined by patellar tracking path, which was guided by engagement between patella and trochlear groove, as mentioned above [205]. For TKR-inserted knee joint, since the articular congruence varies upon the specific implant design and differs from the intact condition, patellar motion magnitudes and variation pattern could be inconsistent among cases.

Merican et al. further specified patellar lateral tilt for TKR-inserted knee joint utilizing cadaver specimens installed onto a custom made jig. Excessive amount of lateral tilt was observed from 10° to 45° flexion, and the maximum tilt magnitude approached 15° around 45° flexion [201]. Comparatively, tilt angle measured from the neutral case (before LRR procedure) in the current study showed similar curve variation trend and close magnitude during knee joint motion, yet the peak value occurred at even lower flexion angle (20°). This finding could relate to the specific intraoperative condition the current study simulated. At the lower knee flexion angles, joint fit between the patella and trochlear groove was not fully formed, so patellar kinematics was majorly controlled by surrounding soft tissue, especially the mutual balance between the medial and lateral patellofemoral retinacula. Within the intraoperative condition, after TKR insertion procedure with incision at the medial side, the knee capsule was loosely sutured back to ease the trial process during surgery for selecting appropriate patellar button. Therefore, during knee joint flexion/extension, unbalance between the medial and lateral compartment in addition to difference in material property tended to be intensified due to
physical inefficiency of the medial retinaculum, which could have further led to excessive patellar lateral tilt.

Hsu et al. and Merican et al. clearly identified that patellar thickness increase directly correlates with excessive patellar lateral tilt [65, 201]. Their conclusion is well consistent with the observation from the current study, as demonstrated by Figure 7.5a and 6a. Furthermore, based on a specific postoperative study, Youm et al. also found that patellar lateral tilt was greater in patients with patella more than 1 mm thicker than its original thickness after resurfacing procedure [126]. From a biomechanical perspective, the medial patellofemoral retinaculum, specifically the medial patellofemoral ligament is substantially thinner, less stiff and stronger than the fibrous structures connecting the patella to the iliotibial band [206, 207]. When patellar thickness increases, the lateral retinacular structure is more prone to be over-tensed during knee joint motion, which could resultantly lift the patellar medial pole and compress the lateral edge contact against the trochlear groove, causing ultimately excessive lateral tilt.

Clinically, excessive patellar tilt could alter the contact pattern within the PF joint during patellar tracking, causing potential patellar maltracking, subluxation or dislocation. Also, lateral tilt has been considered a contributor to the abnormal lateral PF joint contact pressure [65]. In the current study, lateral retinacular release procedure significantly reduced lateral tilt, also supported by Merican et al [201].

As mentioned previously, previous studies [65, 205] recognized that after a short stage of medial shift, gradual increase of lateral shift occurred during intact patellar tracking from 20° to approximately 90° flexion. This motion is closely related with
trochlear geometry and based on the mechanism as follows: before full engagement of patella into the trochlear groove, patella slides into the trochlear groove from its proximal-medial edge guided by tensed medial patellofemoral ligament, and as the knee flexion angle progresses, the patella is guided specifically following the central axis of the femoral trochlear groove, which inclines distally and laterally [205]. In contrast, for the TKR-inserted knee in the current study, standard domed-shaped patellar button was used, and tight geometrical congruence of patellotrochlear articulation was not acheived, which can be observed from the concentrated pressure distribution reported in Figure 7.9. Patellar M/L shift was determined by less-conformal TKR geometry and interaction between the medial and lateral soft tissue compartments. Considering the aforementioned intraoperative condition, observations presented in Figure 7.5c and 6c can be reasonably accepted. Besides, since the LRR procedure relaxed the lateral retinaculum, reduction of the lateral shift magnitude can be well expected.

Among all components of patellar motion reported in the current study, the mechanical significance of patellar M/L rotation remained unknown. From previous measurement, small amount of patellar rotation within the range of 3-5° [65, 205, 208] was recorded, and results obtained from the current study fell into this range as well. Interestingly, similar curve patterns seen in Figure 7.5b and Figure 7.6b were also found in patellar lateral rotation curves for intact knee specimens.

It is well-accepted that patellar flexion angles changes following a monotonic trend in TKR-inserted knee joints but lags behind by approximately 30% compared with tibiofemoral flexion [65, 209]. Maximum patellar flexion angle reached in the current
study was close to that obtained from literatures. As noticed in the current project, patellar thickness alteration demonstrated significant impact on patellar flexion angles. A possible explanation for this observation could be associated with patellar posture. For thicker and bulky patella with greater inertia effect, tensed quadriceps tendon (especially the distal portion connecting the patellar apex to tibial tuberosity) which drives patellar tracking along the arc could cause patellar hinge effect around its contacting point with trochlea articulation. This led to additional patellar inclination (i.e. flexion) about the femoral medial-lateral axis, and can be associated with the larger contact area seen slightly distal from the center of articular surface, as indicated in Figure 7.9c and f.

Using specimen-specific computational FE analysis for TKR-inserted knee, Fitzpatrick et al. varied the patellar resection thickness as well as overall thickness of the composite at 4 levels and found that the thicker patella reduces the requirement of quadriceps tendon force during lower or intermediate flexion due to addition of moment arm of the extensor mechanism, while increases the corresponding magnitude at higher flexion, which could be a result of change in wrapped vastus medialis and vastus lateralis bundles around the femoral component [138]. Consistent finding was also obtained from the current study. Regarding the variation pattern of the quadriceps tendon force within an individual case, as knee flexion angle increases, quadriceps tendon is gradually and passively stretched, thus leading to higher internal force reaction [210]. A notable value drop around 70° flexion could be associated with unsmooth transition the patella underwent from tracking along trochlear groove to artificial femoral condyles as the knee joint proceeded to deeper flexion angles [209], since sudden radius change of the arc
(femoral condylar profile) patella tracks along with could vary the passive tension within the quadriceps tendon.

In the current study, the PF contact pressure reached a maximum around 60° knee flexion, rather than at deep flexion angles. Interestingly, the similar pattern was observed in a few previous studies done by Huberti et al. and Seedholm et al. for intact patellofemoral joint [43, 211], in which they claimed peak PF contact pressure was found at 60° and 90° flexion. According to Hertz theory, the contact pressure value depends not only on patellofemoral compressive force, but also the mutual patellotrochlear geometrical fit for contact area. As patellar thickness advances, as observed from the current study, excessive patellofemoral contact pressure was detected, which is consistent with Hsu et al [65]. The extra PF contact pressure could also be an indicator of the overstuffed patellofemoral joint, as mentioned in the introduction section. The LRR procedure could effectively reduce tension within the lateral retinacular structure, by which the resultant force component along the patellar anterior-posterior direction for compressing patella towards trochlear groove could be reduced as well. Thus reduced PF contact pressure after performing LRR procedure could be well anticipated.

The current project also arose a few limitations. It was assumed that patella followed unique and same paths of motion between knee flexion and extension. However, Amis et al. noted that intact patella followed different paths during knee flexion and extension, even though they could not confirm a statistical significance [205]. Based on this standpoint, it could be difficult to conclude that our testing measurements from knee extension motion fully reproduced knee flexion motion. Another limitation is
about processing the motion tracking data. The origin of motion (zero position) was highly dependent upon the experimental conditions and specimen geometry. Many researchers utilized customized positions to zero all rotational and translational motion, and in the current study, the full extension state was selected as the reference position/orientation. However, this reference may numerically impose unnecessary constraints to patellar motion [204]. Also, pre-existing motion might be available at the extension state especially if alignment inaccuracy existed. For calibration purpose, extensive use of imaging measurement could be desirable to inspect the actual patellar initial position and orientation relative to the patellar and femoral body-fixed axes [204]. Finally, to obtain a generic conclusion regarding the effect of patellar component thickness on patellar kinematics and patellofemoral joint function, it would be essential to recruit more cadaver specimens from various patient groups for more generalized and comprehensive study.

7.4 Conclusions

The current study utilized patient-specific cadaver study to explore the intraoperative effect of patellar component thickness on patellar kinematics and patellofemoral joint mechanics. Optical motion tracking system and customized UHMEPE pressure sensor were also implemented for quantitative measurement. Clinically-prevalent lateral retinaculum release procedure was also performed for comparative testing results. It was found that patellar thickness increase causes extra amount of patellar lateral tilt, shift and patellar flexion. Thicker patella may contribute to decreased quadriceps tendon force at lower flexion angles, but also leads to amplified PF
joint contact pressure. Also, LRR procedure could significantly enhance patellofemoral joint function by reducing lateral tilt and shift, as well as PF contact pressure magnitude.

7.5 Acknowledgement

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CHAPTER 8  CONCLUSIONS

This project consolidated the use of the computational simulation and experimental exploration, and provided a good reference for relevant studies on interactively utilizing advantages of both methods to address topics. The computational study provided a straightforward prediction for possible experimental outcomes, while the experimental study considered more realistic surgical condition and further extended the understanding regarding patellofemoral joint mechanics. Results obtained from both the computational simulation and cadaver experiment showed great agreement on the effect of patellar component thickness on quadriceps tendon force and patellofemoral contact pressure, which further demonstrated the feasibility of the research method and testing platform developed in the current project.

From the perspective of clinical application, this study specifically addressed the biomechanics issues associated with the patellar resurfacing procedure, to reduce the risk caused by various complications and provide assistance for practitioners during the operation in implant selection. Also, Stryker Triathlon® TKR device, which was release in 2005, has become currently one of the most popular TKR implant in use. However, there are few specific studies considering the functionality and potential complications regarding this device, so this research project made beneficial contribution to the current literature, and could further support improvement of the implant design.

In general, patellar thickness variation significantly affects the patellar kinematics, quadriceps tendon force and patellofemoral contact pressure. Extensive patellar lateral tilt,
shift and patellar flexion could be expected with the patellar thickness increase. At lower flexion angles, thicker patella tends to be beneficial in reducing the quadriceps tendon force, however, at higher flexion angles, overly thick patella could lead to extra quadriceps tendon force. Thinner patella contributes to inefficiency of extensor mechanism, which means, higher quadriceps tendon force is well anticipated during the normal knee joint motion cycle. Nevertheless, reduced patellar thickness is good for releasing patellofemoral joint by decreasing the contact pressure.
CHAPTER 9 RECOMMENDATIONS FOR FUTURE WORK

To obtain more advanced understanding about patellar biomechanics, and find a practical solution to determine the optimum patellar component during TKR surgery, a few recommendations are provided:

**a.** In the current project, more cadaver specimens should be involved in the in-vitro study for statistical significance. In addition, due to anatomical difference between the specimens used in the FE model and cadaver study, convergence problem was noticed regarding data interchange between two studies. To address this issue, same specimens should be recruited throughout the project, and initial position of TKR components relative to motion tracking markers should be also recorded for reference.

**b.** Enhanced computational models are expected, which are defined based on MRI imaging of cadaver specimens for realistic geometry, in-vivo or in-vitro loading/kinematics inputs, and validated by cadaver testing or clinical trials with sufficient statistical significance.

**c.** A finite element model defined based on postoperative condition should also be considered for comprehensive evaluation. Advanced constitutive material model should be specified for active muscular contraction (eccentric, concentric and isometric).

**d.** Postoperative experimental testing based on physical therapy practice should be comparatively used with the finite element model.
e. Pressure sensor with smaller sensing units, improved mapping resolution and wireless data communication should be further developed for more robust gauging performance in realistic clinical applications.

f. Interactive graphic user interface of the pressure sensor should be built compatible with Apple or Android system to enable use of the pressure sensor in mobile devices for better usability.
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