REDUCING MOBILE BEARING DISLOCATIONS IN THE OXFORD DOMED LATERAL UNICOMPARTMENTAL KNEE REPLACEMENT IMPLANT

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ABSTRACT

The Oxford Domed Lateral (ODL) mobile bearing Unicompartmental Knee Replacement (UKR) provides good clinical results, but the bearing dislocation rate is unacceptably high.

A systematic review and meta-analysis found that changing the articulating surface on the tibial component from a flat to a domed shape reduced the overall bearing dislocation rate from 17% to 3.7%. The risk of anterior/posterior (AP) dislocations was reduced substantially. Most dislocations were found to occur medially and with the knee in a flexed position.

Dislocations occur in flexion because when the knee is flexed, the lateral ligaments are lax. Therefore a mechanical rig, which allowed the relative position of components to be adjusted and the joint to be distracted, was used to determine the minimum Distraction required for a Dislocation (DD), which is related to the risk of dislocation. However, as testing with the rig was inefficient, a computational dislocation analysis tool based on the mechanical rig was developed and validated against results from the rig. The tool uses path planning algorithms from the field of robotics to allow the bearing to find a viable dislocation pathway. It was found that medial dislocations required a much lower DD than AP dislocations, which explains why medial dislocations are more common than AP dislocations. Further, changing the tibial surface from a flat to a dome increased the DD for AP dislocations substantially, which explains why this change primarily decreased AP dislocations.

A series of potential implant design changes were studied to see if they would decrease the DD for medial dislocations to that of AP dislocations as the risk of AP dislocation was considered acceptable. The simplest change to achieve this was to increase the tibial wall height by 2 mm, but to avoid the femoral and tibial components hitting if there is bearing wear, the minimum bearing thickness would have to be 5 mm. Alternatively if the centre of the femoral component sphere was moved laterally, the bearing made wider laterally and the wall height increased, then an acceptable dislocation rate would be achieved with a minimum bearing thickness of 4 mm. For both these modifications to be successful, surgeons must ensure that the bearing is not more than 4 mm from the tibial wall.

Cadaveric and clinical studies are required to assess the viability of the proposed changes. If the risk of dislocation can be minimised or reduced, more patients may benefit from the advantages of lateral mobile bearing UKRs.

STATEMENT OF CONTRIBUTION

This thesis is submitted in accordance with the rules laid down by the Medical Sciences Division and the Graduate Studies Department of The University of Oxford. It is submitted in partial fulfilment of the requirements for award of the degree of Doctor of Philosophy in Musculoskeletal Sciences.

The work presented in this thesis is Irene Yang's work except where due reference or acknowledgement is given to the work of others.

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DEDICATION

This thesis is dedicated to my family:

My other half, Russell Buchanan,

My mother 洪明秀 and father 杨叶平,

Jen, Nyan and my niece and nephew, Avalon and Hunter Khin,

to 杨祥锥, 杨如光 and Margaret, 袁立

And finally, to my family members who have passed.

TABLE OF CONTENTS

Abstract i

Statement of contribution ii

Copyright Statement iii

Acknowledgements iv

Dedication v

Table of Contents vi

List of Abbreviations xiv

List of Figures xvi

List of Tables xxi

Introduction xxiv

CHAPTER 1 LITERATURE REVIEW 1

1.1 The knee joint 1

1.1.1 Knee joint stabilisers: Ligaments and muscles of the knee 3

1.1.2 The lateral knee 7

1.1.2.1 The lateral femoral and tibial condyles 7

1.1.2.2 The lateral meniscus 7

1.1.2.3 Physiology of the lateral knee 9

1.2 Osteoarthritis (OA) of the knee 11

1.2.1 Prevalence and aetiology of lateral compartment knee OA 11

1.2.2 Diagnosis 13

1.2.3 Treatment 13

1.2.4 Conservative treatment/non-operative 13

1.2.5 Surgery 15

1.3 Lateral unicompartmental knee replacement 17

1.3.1 LUKR Implants 26

1.3.2 Fixed versus mobile bearing LUKR 30

1.3.3 The Oxford Knee implant 31

1.4 The Oxford Domed Lateral (ODL) implant 32

1.4.1 ODL implant: Surgical indications and the procedure 33

1.4.2 ODL implant complications: Mobile bearing dislocation 33

1.4.2.1 Dislocation direction 35

1.4.2.2 Cause of dislocation 36

1.4.2.3 Treatment of dislocations 37

CHAPTER 2 BEARING DISLOCATION IN THE ODL IMPLANT –META-ANALYSIS COMPARING THE OXFORD FLAT LATERAL VS. THE ODL IMPLANT 38

2.1 Introduction 38

2.2 Methods 38

2.2.1 Search strategy 40

2.2.2 Selection criteria for studies 41

2.2.3 Data collection 41

2.2.3.1 Number of patients followed up 42

2.2.4 Strategy for data synthesis 42

2.2.4.1 Statistical analysis *42*

2.2.4.2 Confidence interval *43*

2.2.5 Risk of bias assessment 43

2.3 Results 44

2.4 Discussion 53

2.4.1 Subgroups or subset analyses: deviations from PROSPERO 54

2.4.2 Dislocation and revision rates 54

2.4.3 Additional information on dislocations 55

2.4.4 Study limitations 55

2.5 Conclusions 56

2.6 Acknowledgements 57

CHAPTER 3 THE EFFECT OF COMPONENT POSITION ON THE RISK OF BEARING DISLOCATION: A MECHANICAL STUDY WITH A CUSTOM-DESIGNED RIG 58

3.1 Introduction 58

3.2 Aims 60

3.3 Chapter 3 Part A: Modifying the mechanical rig to mimic knee flexion 61

3.3.1 Aims 64

3.3.2 Methods 64

3.3.2.1 Modifying the rig to mimic knee flexion *64*

3.3.2.2 Validating the modified rig results: comparison with Weston-Simons et al.'s study *66*

3.3.3 Results 68

3.3.3.1 Modifying the rig to mimic knee flexion 68

3.3.3.2 Validating the modified rig results with Western-Simons et al. 69

3.3.4 Discussion 70

3.3.5 Conclusion 74

3.4 Chapter 3 Part B: The effect of component position on the risk of bearing dislocation 76

3.4.1 Aims 76

3.4.2 Methods 76

3.4.2.1 Defining "dislocation" 76

3.4.2.2 Testing dislocations using the modified mechanical rig 76

3.4.2.3 Validating the results with an independent user: Inter-observer assessment using Inter-class Correlation Coefficient (ICC) 77

3.4.3 Results 78

3.4.3.1 Defining "dislocation" 78

3.4.3.2 Testing dislocations using the modified rig 80

3.4.3.3 Validating the results with an independent user: Inter-observer assessment using Inter-class Correlation Coefficient (ICC) 83

3.4.4 Discussion 85

3.4.5 Study limitations 90

3.4.6 Conclusion 91

3.5 Chapter 3 Discussion and conclusions 92

CHAPTER 4 A COMPUTATIONAL MODEL OF MOBILE BEARING DISLOCATION USING ROBOTICS PATH PLANNING ALGORITHMS 93

4.1 Introduction 93

4.2 Aims 95

4.3 Methodology 96

4.3.1 The setup 96

4.3.2 Implementing the "smart search" 97

4.3.3 Selecting search parameters 98

4.3.4 Testing 98

4.4 Results 99

4.4.1 Selecting optimal search parameters 100

4.4.2 Validation of dislocation results 101

4.5 Discussion 102

4.5.1 Study limitations 106

4.6 Conclusions 106

CHAPTER 5 COMPARISON OF DISLOCATION RISK TO DISLOCATION RATES: FLAT VERSUS DOMED OXFORD IMPLANT 107

5.1 Introduction 107

5.2 Aims 107

5.3 Methods 108

5.3.1 Dislocation rates calculated from literature 108

5.3.2 Dislocation risk calculated computationally 108

5.4 Results 109

5.4.1 Dislocation rates by dislocation direction 109

5.4.2 Dislocation risk calculated computationally 112

5.5 Discussion 113

5.5.1 Dislocation rates by dislocation direction 113

5.5.2 Dislocation risk calculated computationally 114

5.5.2.1 Dislocation risk calculated computationally: comparison with Weston-Simons et al. 115

5.5.3 Dislocation rate versus dislocation risk 116

5.5.4 Study limitations 118

5.6 Conclusion 118

CHAPTER 6 DESIGN MODIFICATIONS TO THE ODL IMPLANT TO REDUCE MEDIAL BEARING DISLOCATIONS 119

6.1 Introduction 119

6.2 Aims 119

6.3 Methods 119

6.4 Results 120

6.5 Discussion 133

6.6 Conclusions 134

CHAPTER 7 THE NEW ODL IMPLANT DESIGN TO RESIST MEDIAL DISLOCATIONS – A PROPOSAL 135

7.1 Introduction....135

7.2 Aims....135

7.3 Methods....136

7.3.1 Designing and modelling the modified components....136

7.3.2 Testing the modified components using the dislocation analysis tool....137

7.3.3 Validating the computational results using the mechanical rig....139

7.4 Results....141

7.4.1 Higher tibial wall results....141

7.4.2 Tibial dome shift results....145

7.4.3 Wider femoral component results....146

7.4.4 Femoral dome shift results....147

7.4.5 Wider bearing results....148

7.4.6 New design results....149

7.4.7 Validation using the mechanical rig....150

7.5 Discussion....151

7.6 Conclusions....154

7.7 Acknowledgements....155

CHAPTER 8 THESIS CONCLUSIONS AND SUGGESTIONS FOR FUTURE WORK....156

8.1 Thesis conclusions....156

8.1 Overall thesis limitations 160

8.2 Future work 161

8.3 Conclusion 163

References 164

Appendix A: MATLAB computational tool to assess bearing dislocation 180

Appendix B: Robotics dislocation analysis code 188

Appendix C: Relevant research output 190

LIST OF ABBREVIATIONS

ACL	Anterior Cruciate Ligament
AKSS	American Knee Society Score
AP	Anterior-Posterior
BKS	Bristol Knee Society
BMI	Body Mass Index
CAD	Computer Aided Design
CI	Confidence Interval
DD	Distraction to Dislocation
DFO	Distal Femoral Osteotomy
FLO	Fixed Lateral Oxford
GA	Greta Agustoni
GUI	Graphical User Interface
HSS	Hospital for Special Surgery
HTO	High Tibial Osteotomy
IBME	Institute of Biomedical and Mechanical Engineering
ICC	Intra-class Correlation Coefficient
IKSS	International Knee Society Score
IY	Irene Yang
JOAKS	Japanese Orthopaedic Association Knee Score
KOOS	Knee Injury and Osteoarthritis Outcome Score
KOOS, JR	Knee Injury and Osteoarthritis Outcome Score, Junior
KSS	Knee Society Score
LCL	Lateral Collateral Ligament
LCS	Low contact Stress
LUKR	Lateral Unicompartmental Knee Replacement
MCL	Medial Collateral Ligament
MINORS	Methodological Index for evaluation of Non-Randomized Studies
ML	Mediolateral

NICE	National Institute for Health and Care Excellence
NJR	National Joint Registry
NSAIDs	Non-Steroidal Anti-Inflammatory Drugs
OA	Osteoarthritis
ODL	Oxford Domed Lateral
OKS	Oxford Knee Score
OMPL	Open Motion Planning Library
ORS	Osteoarthritis Research Society
PCL	Posterior Cruciate Ligament
PICOS	Population, Intervention, Comparators, Outcomes and Studies
PCA	Partial Condylar Arthroplasty
PKA	Partial Knee Arthroplasty
PKR	Partial Knee Replacement
PLA	Polylactic Acid
PMMA	Polymethylmethacrylate
PTIR	Patient Time Incidence Rate
RCT	Randomised Control Trial
RRT	Rapidly-exploring Random Trees
SD	Standard Deviation
SF-36	Short Form/SF-36
TKR	Total Knee Replacement
UCA	Unicondylar Arthroplasty
UCR	Unicondylar Replacement
UKR	Unicompartmental Knee Replacement
VAS	Visual Analogue Scale
WOMAC	Western Ontario and McMaster Universities Arthritis Index

LIST OF FIGURES

Figure 1.1. Anterior and posterior view of the right human femur (left) and tibia (right) [15]. 2

Figure 1.2. Posterior view (left) and side view (right) of the major ligaments in the right knee. 4

Figure 1.3. Muscles of the anterior right thigh. 5

Figure 1.4. Muscles of the posterior right thigh. 5

Figure 1.5. Muscles of the posterior right lower leg. 6

Figure 1.6. Topographical presentation of the articular surface of the right leg showing proximal tibial condyles (A) and the distal femoral condyles (B) [14]. In image A the medial condyle is on the left, and in image B the medial condyle is on the right. 7

Figure 1.7. Superior view of the medial and lateral menisci on the right tibia. 8

Figure 1.8. Posterior knee showing the ligamentous connections of the posterior horn of the lateral meniscus [25]. 9

Figure 1.9. The relative movement of femoral condyles from extension to 120° flexion (right knee) [29]. 10

Figure 1.10. The centre of rotation of the femur lies posterior to the tibial plateau during high flexion (162° depicted) [28]. 11

Figure 1.11. UKR (left) vs. TKR (right) X-ray. 16

Figure 1.12. The Oxford Knee. The medial implant features a flat tibial surface [156]. 31

Figure 1.13. The Oxford Domed Lateral implant [113]. 32

Figure 1.14. Screws inserted into the tibial eminence to discourage medial dislocation of the mobile bearing [8]. 34

Figure 1.15. Annotated X-Ray of a patients left knee. Arrows indicate the radio-opaque markers and the dislocated bearing is outlined in red. 35

Figure 2.1. PRISMA flow diagram for study inclusion. 45

Figure 2.2. Time of first dislocation(s) with respect to mean follow up duration in months. The mean time of first dislocation (16 months) is marked and labelled. 49

Figure 2.3. Percentage bearing dislocation assessment of flat versus domed tibial component design. 51

Figure 2.4. Percentage bearing dislocation assessment of flat versus domed tibial component design – excluding Kehr et al. study. 51

Figure 2.5. PTIR assessment for all cause revision: Flat versus domed tibial component. 52

Figure 2.6. PTIR assessment of bearing dislocation: flat versus domed tibial component design. . 52

Figure 2.7. PTIR assessment for all cause revision except bearing dislocation: Flat versus domed tibial component..... 53

Figure 2.8. Percentage bearing dislocation: designer versus non-designer..... 53

Figure 3.1. Mechanical rig model the Oxford Domed Lateral implant femoral component fixed in neutral, a domed tibial component and 3 mm mobile bearing [10]. 58

Figure 3.2. Schematic depicting how knee flexion can result in up to 50% of the bearing overhanging the tibial component posteriorly (relative to the right knee). 59

Figure 3.3. Bearing overhang (mm) (relative to the tibial component) versus flexion angle (°) for the ODL implant [158]..... 60

Figure 3.4. Metal calibration plates of thicknesses 2 mm, 3 mm, 4 mm, 5 mm and 6 mm. 61

Figure 3.5. Scatter plot of plate thickness (mm) versus measured voltage (V). 62

Figure 3.6. The starting position showing the bearing flush against the wall of the tibial component and with the femoral component resting on the superior surface of the bearing. 66

Figure 3.7. The minimum distance of the femoral component to tibial wall “a” and the minimum distance from the femoral component to the domed tibial surface “b”..... 67

Figure 3.8. Rotating 75 mm arc stage platform with top and bottom halves. 68

Figure 3.9. Final setup of the mechanical rig for testing. Note the spring in compression on the left hand side. The modified bearing with a handle attached is shown in the picture. 69

Figure 3.10. Schematic representing level 1, level 2 and level 3 medial dislocation with the Oxford Domed Lateral implant. 79

Figure 3.11. Schematic representing femoral congruency, tibial congruency and non-congruent dislocations for lateral and anterior/posterior dislocation with the Oxford Domed Lateral implant. 80

Figure 3.12. The effect of tibial rotation on medial dislocation: Central bearing with no overhang, 1 mm, 10 mm and 17 mm posterior bearing overhang. 82

Figure 3.13. The effect of posterior bearing overhang on medial dislocation at 0°, 10° internal, 30° internal, 10° external and 30° external tibial rotation angles. 82

Figure 3.14. Mechanical rig results comparison between IY and GA testing: medial dislocation, lateral dislocation and anterior/posterior dislocation. 84

Figure 3.15. Geometric representation of the maximum amount of displacement unaccounted for in testing by not zeroing between readings. Note: this figure is not drawn to scale. 88

Figure 4.1. Projection of RRT and the optimal computed path for a 2D rigid robot body to navigate in a 2D environment space from a starting to an ending point. Note the obstacles shown in black [182]. 95

Figure 4.2. Bearing (teal colour) dislocations with bounding box (white outline) and goal area (red outline): (a) medially, (b) laterally, (c) anteriorly or (d) posteriorly. The top panel shows top view and the side view is shown in the bottom panel. 97

Figure 4.3. The effect of number of searches (n) on solution identification. 100

Figure 4.4. The effect of search time (s) on solution identification. 100

Figure 4.5. Comparing the mobile bearing dislocation (medially, laterally, anteriorly or posteriorly) results obtained using the robotics dislocation analysis tool versus mechanical rig. The grey region marks the dislocation zone and the white region marks the no dislocation zone... 101

Figure 4.6. Bland-Altmann assessment for medial dislocation comparing mechanical rig vs. robotics dislocation analysis tool. 102

Figure 5.1. Flow chart showing clinical dislocation rates for dislocation directions, assuming dislocations with unspecified directions were antero-posterior. 111

Figure 5.2. Medial, anterior or posterior dislocation: Oxford flat lateral versus ODL implant. 112

Figure 5.3. The effect of changing the tibial component from a flat to a domed tibial surface. Line 1 is located in the centre of the bearing at the peak of the dome, line 2 and line 3 are located on the medial side of the bearing, where line 2 aligns with the vertical centre and line 3 aligns with the maximum heights located at the anterior/posterior extremities. 116

Figure 5.4. Flat (left) versus domed (right) tibial component wall. 117

Figure 7.1. Series of images depicting 0 mm, 2 mm, 4 mm and 6 mm ML..... 138

Figure 7.2. The new femoral, tibial and mobile bearing components in the mechanical rig. 139

Figure 7.3. The effect of bearing thickness and increased tibial wall height on anterior and posterior dislocation. NOTE: Current design refers to the 3 mm bearing with standard tibial component wall height. 141

Figure 7.4. The effect of bearing thickness on medial dislocation. NOTE: Current design refers to the 3 mm bearing with standard tibial component wall height. 142

Figure 7.5. The effect of increasing wall height and using the corresponding thinnest bearing possible on medial dislocation. NOTE: “Current design” refers to the 3 mm bearing with standard tibial component wall height. There is no 7 mm line plotted because in the parameters tested, the 7 mm bearing did not dislocate. 143

Figure 7.6. Straight line showing the smallest gap distance between the femoral sphere (representing the femoral component) and the tibial component. 144

Figure 7.7. Shifting the centre of the sphere defining the tibial component either medially (towards the tibial wall) or laterally (away from the tibial wall). 145

Figure 7.8. Increasing the width of the femoral component either medially (towards the tibial wall) or laterally (away from the tibial wall). 146

Figure 7.9. Shifting the centre of the sphere defining the femoral component either medially (towards the tibial wall) or laterally (away from the tibial wall). 147

Figure 7.10. Increasing the bearing width either medially/ towards the tibial wall (left) or laterally/ away from the tibial wall (right). NOTE: Zero marks = starting point where the bearing is flush against the tibial wall if the components were unaltered. The ML distance is relative to the zero/start position. 148

Figure 7.11. New design - one medial dislocation solution depicted in top view and side view... 149

Figure 7.12. Combined design modifications to the ODL implant. NOTE: minimum bearing thickness of 4 mm. Red line shows the minimum DD for anterior/posterior dislocation at 5.5 mm. 149

Figure 7.13. Validation of the results for the new design of the ODL implant comparing results obtained using the mechanical rig and the dislocation analysis tool..... 150

LIST OF TABLES

Table 1.1. Major ligaments of the knee joint. 4

Table 1.2. Muscles that control knee movement [15]. 5

Table 1.3. Motion in the knee joint [13]. 9

Table 1.4. Knee scores used to measure LUKR functional outcomes 18

Table 1.5. LUKR clinical outcomes and implant survival rates. 19

Table 1.6. Fixed bearing LUKR implants reported in literature and/ or registries. 27

Table 1.7. Mobile bearing LUKR implants reported in literature and/ or registries. 29

Table 1.8. LUKR implants with mobile/fixed bearing options reported in literature and/ or registries. 29

Table 2.1. PICOS outline to define the systematic review question. 39

Table 2.2. Database search results. 44

Table 2.3. Excluded studies with reasoning after reading full manuscript. 46

Table 2.4. Risk of bias assessment for included studies. 47

Table 2.5. Demographic information of studies included in quantitative analysis. 48

Table 2.6. Dislocation with patient specific information available. 48

Table 2.7. Mobile bearing dislocation data for all studies reporting no. of dislocations. 50

Table 3.1. Voltage output readings for the known plate thicknesses. 61

Table 3.2. Modifications made to the mechanical rig. 63

Table 3.3. Modified mobile bearings designed. The bearing without any tabs is the bearing used for testing. The remaining bearings are used to setup overhang testing - red shows where the tabs were placed. 65

Table 3.4. Weston-Simons et al.'s validation testing parameters. 66

Table 3.5. Validation testing results comparing the results from Weston-Simons et al. using the original rig with the dislocation results measured in this study using the modified rig. 70

Table 3.6. Repeating Weston-Simons et al. [10] using swing dislocation manoeuvre. 72

Table 3.7. New dislocation testing parameters. 76

Table 3.8. ICC Validation testing parameters. 77

Table 3.9. The minimum DD (mm) required for medial, lateral, anterior or posterior dislocation of the mobile bearing given various configurations of the metal femoral and tibial components. 80

Table 3.10. GA results: amount of DD (mm) required for medial dislocation of the mobile bearing given various configurations of the metal femoral and tibial components. 85

Table 4.1. Top: Dislocation results obtained using the mechanical rig vs. the robotics dislocation analysis tool. 101

Table 5.1. Clinical dislocation rates by dislocation direction: flat tibial component design..... 109

Table 5.2. Clinical dislocation rates by dislocation mode: domed tibial component design. 110

Table 5.3. Comparing the current results with Weston-Simons et al. [10]: Distraction to Dislocation (DD) medially or anteriorly/posteriorly using either the Oxford flat lateral implant or the Oxford Domed Lateral (ODL) implant measured when the bearing is 1 mm from the tibial wall. 115

Table 6.1. User needs. 121

Table 6.2. ODL implant components: features to alter/target for design changes to reduce mobile bearing dislocation. 122

Table 6.3. Design ideas to reduce dislocation of the mobile bearing. 123

Table 6.4. Final tibial component modifications selected for testing. NOTE: components depicted with the knee in flexion. 130

Table 6.5. Final femoral component modifications selected for testing. NOTE: components depicted with the knee in flexion. 131

Table 6.6. Final mobile bearing component modifications selected for testing. NOTE: components depicted with the knee in flexion. 132

Table 7.1. Testing parameters. 140

Table 7.2. Tibial wall adjustments based on femoral dome shift. 147

INTRODUCTION

Knee osteoarthritis (OA) is a debilitating musculoskeletal disorder characterised by deterioration of the articular cartilage lining the ends of bones and subchondral bone causing bone-on-bone articulation leading to pain, compromised function and ultimately joint destruction [1]. The global prevalence of knee OA is 16% in people aged 15 and over, which increases to 22.9% in those aged 40 and over [2]. Isolated lateral knee OA is less common than medial knee OA [3], accounting for 5 - 10% of arthritic knees [3, 4] and affecting 32.7 million people globally over the age of 40 [2, 3].

Traditionally, patients with end stage lateral knee OA are treated through surgery. Knee replacement surgeries can be broadly divided into Total Knee Replacement (TKR) or Unicompartmental Knee Replacement (UKR) surgeries. UKR further comprises medial UKR or lateral UKR depending on the knee compartment (either medial or lateral) affected by OA. For lateral UKR, the Oxford Domed Lateral (ODL) UKR implant is the only mobile bearing implant designed specifically for use laterally [5]. Clinically, the ODL performs well [6], however, the results have been marred by the problem of dislocation of the mobile bearing, with the bearing typically lodging at an angle on top of the medial wall of the tibial component. In the literature, clinical journal articles report dislocations as a percentage dislocation rate. In line with this, the term “dislocation rate” has been used in this thesis to refer to the incidences of dislocations that have occurred in clinically reported patient groups reported in the literature. On the other hand, the term “dislocation risk” has been used to refer to the chance of sustaining a bearing dislocation – of the total configurations tested, implant designs that had more configurations in which dislocation was possible had a greater dislocation risk than designs that had less configurations that could result in bearing dislocation. However, despite attempts to reduce the dislocation risk [7, 8], clinical dislocation rates remain unacceptably high (1 - 6%) [9]. The aim of this thesis was firstly to gain a better understanding of the problem of mobile bearing dislocation and secondly, to reduce the risk of medial mobile bearing dislocation by improving the design of the implant. This thesis also provides surgical recommendations to reduce the risk of medial bearing dislocations.

Chapter 1 provides background information starting with an overview of anatomy of the lateral knee, lateral knee osteoarthritis and its treatments. It encompasses an overview of the performance

of the lateral knee implant, a review of the information available on the problem of mobile bearing dislocation, as well as the attempts that have been made to reduce dislocations.

Chapter 2 presents the results of a systematic review which aimed to understand the effect of modifying the design of the implant on bearing dislocation: A meta-analysis was conducted to explore the effect of changing the tibial component from a flat to a domed shape on the clinical rate of bearing dislocation.

With a better understanding of the literature regarding the ODL implant and mobile bearing dislocation, **Chapter 3 Part A** builds on the work first published by Weston-Simons et al. [10] on the use of a mechanical rig designed to study dislocation of the mobile bearing with the Oxford knee implant. The rig allows the user to distract the femoral component away from the tibial component to measure the amount of Distraction that is required for a bearing to Dislocate (DD). The design of the mechanical rig was improved to enable the rig to record DD values more reliably. The rig was also successfully modified to mimic the relative position of the femoral and tibial components when the knee flexes. In **Chapter 3 Part B** the minimum Distraction required for Dislocation (DD) was measured using the modified rig. It was found that medial dislocations required a lower DD than anterior/posterior (AP) dislocations, which partially explains why most clinical dislocations occur medially. Further, the reason bearing dislocations occur in flexion is because knee flexion reduces the tension in the lateral ligaments leading to ligament laxity, which allows the lateral knee to distract. Tibial rotation/posterior bearing overhang were found not to affect the dislocation risk, but the risk of dislocation was found to increase with increasing distance between the bearing and the tibial wall.

Testing dislocation using the mechanical rig was time consuming, prone to bias and posed difficulties when considering testing new implant designs. Therefore, **Chapter 4** describes the development of the robotics dislocation analysis tool. It is built using a modified version of The Open Motion Planning Library (OMPL) software, an open-source path planning library consisting of various state-of-the-art sampling-based motion planning algorithms and custom Graphical User Interface (GUI) [11]. The tool uses path planning algorithms which were developed for use in robotics to solve challenging problems encountered in autonomous motion, to allow the bearing to explore its

environment for a viable dislocation pathway. Using the most common path planning algorithm, the Rapidly-exploring Random Trees (RRT) algorithm, the dislocation analysis tool was used to assess the minimum vertical distraction of the femoral component (relative to the tibial component) required for a mobile bearing dislocation to occur. The Distraction to Dislocation (DD) is inversely related to the risk of bearing dislocation. An additional validation study for the dislocation analysis tool is presented in **Chapter 5** which compares the risk of dislocation to the clinical dislocation rate between the ODL implant and the Oxford flat lateral implant for dislocations anteriorly, posteriorly and medially. It was found that changing the design of the tibial component from a flat to a domed surface reduced the overall risk of dislocation, however, the rate of anterior/posterior dislocation was reduced more than medial dislocation because the design change increased the anterior/posterior bearing entrapment more than the medial entrapment, thereby reducing the risk accordingly.

Chapter 6 explores the results of brainstorming possible modifications that can be made to the ODL implant to reduce the risk of medial bearing dislocation. The chapter presents design modifications for each component and explains the rationale for selecting the most feasible design modifications for further testing. **Chapter 7** presents the results when comparing the bearing dislocation risk (DD medially) between the original ODL implant design and the design modifications selected from Chapter 6. The goal was to design an implant with a DD medially which matched the risk of anterior/posterior dislocation using the original implant components because the rate of anterior/posterior dislocation is clinically acceptable. The chapter concludes with the new implant design which features a combination of design modifications which each successfully reduced the risk of medial dislocation. The new proposed design was then tested against the original implant design. To validate the computational results of the new implant design, the new implant design was also tested using the mechanical rig and the results were compared with the computational results.

The thesis concludes with **Chapter 8**, which summarises the main findings from the thesis, places these findings in the greater context of the literature and outlines recommendations for future research work.

Chapter 1 LITERATURE REVIEW

Some of the work presented below has been submitted as textbook chapter (under peer review at the time of thesis submission)[1].

1.1 The knee joint

The knee joint is one of the largest and most complex joints in the human body. It sustains between 1 to 4.4 times the body weight of individuals, depending on the physical activity e.g. for walking, the body sustains between 2 to 3 times the body weight of an individual [12]. It is composed of two joints, the femorotibial joint and the patellofemoral joint [13, 14] to allow for a highly mobile joint that allows activities such as standing, walking and running [13]. To facilitate frictionless motion, two fibrocartilaginous menisci pads are located between the tibia and the femur, four synovial fluid-filled cavities, termed "bursae", are located in high motion areas of the knee joint to aid movement of tendons and skin over the joint [13]. The knee joint is classed as a synovial joint as the joint is lined on the inner surface of the joint capsule with a synovial membrane, which holds synovial fluid [15]. On the outside of the synovial membrane is a fibrous joint capsule, which protects the thin synovial membrane and provides further structural support to the joint [16].

There are three bones involved in the knee joint: the tibia, the femur, and the patella (see Figure 1.1). The knee joint functions to connect the femur to the tibia allowing transmission of body weight. The femur is the long bone of the upper leg. Distally, the femur flares to form the medial (inner) and lateral (outer) condyles, which are lined with smooth articular cartilage. On each distal femur, is a region medial and lateral to the knee joint not covered in articular cartilage, known as the medial and lateral epicondyle, respectively. The two condyles form the articular surfaces of the femur involved in the knee joint [15].On the anterior surface of the distal femur is the patella bone or kneecap (see Figure 1.1 left, anterior view), the largest sesamoid bone in the human body [15]. The patella is incorporated into the tendon of the quadriceps femoris muscle that crosses the knee joint on the

[1] I. Yang, B. H. van Duren and H. G. Pandit (Submitted), Lateral Unicompartmental Knee Replacement surgery for lateral knee osteoarthritis. Textbook chapter.

anterior surface (see Figure 1.3) [15], and rests in the trochlear groove (patellar surface) (not visible in Figure 1.1 left) [15]. It articulates with the femur in the femoropatellar joint. The patellar has two functions: to reduce friction between the quadriceps femoris tendon and the anterior distal surface of the femur, and to increase the leverage of the quadriceps femoris muscle [15].

The tibia is the main long bone of the lower leg. Proximally, the tibia expands into a large platform, divided into the inner/medial and outer/lateral condyles. Separating the two tibial condyles is the intercondylar eminence. The medial and lateral condyles of the tibia act as receiving platforms for articulation with the femoral condyles in the knee joint [15]. While not formally part of the knee joint, the proximal fibula is an important bone for the attachment of the bicep femoris muscle involved in controlling the knee flexion (see 1.1.1 *Knee joint stabilisers: Ligaments and muscles of the knee*). It is not as robust in structure as the femur or the tibia, and lies parallel and lateral to the tibia (see right image in Figure 1.1) [15].

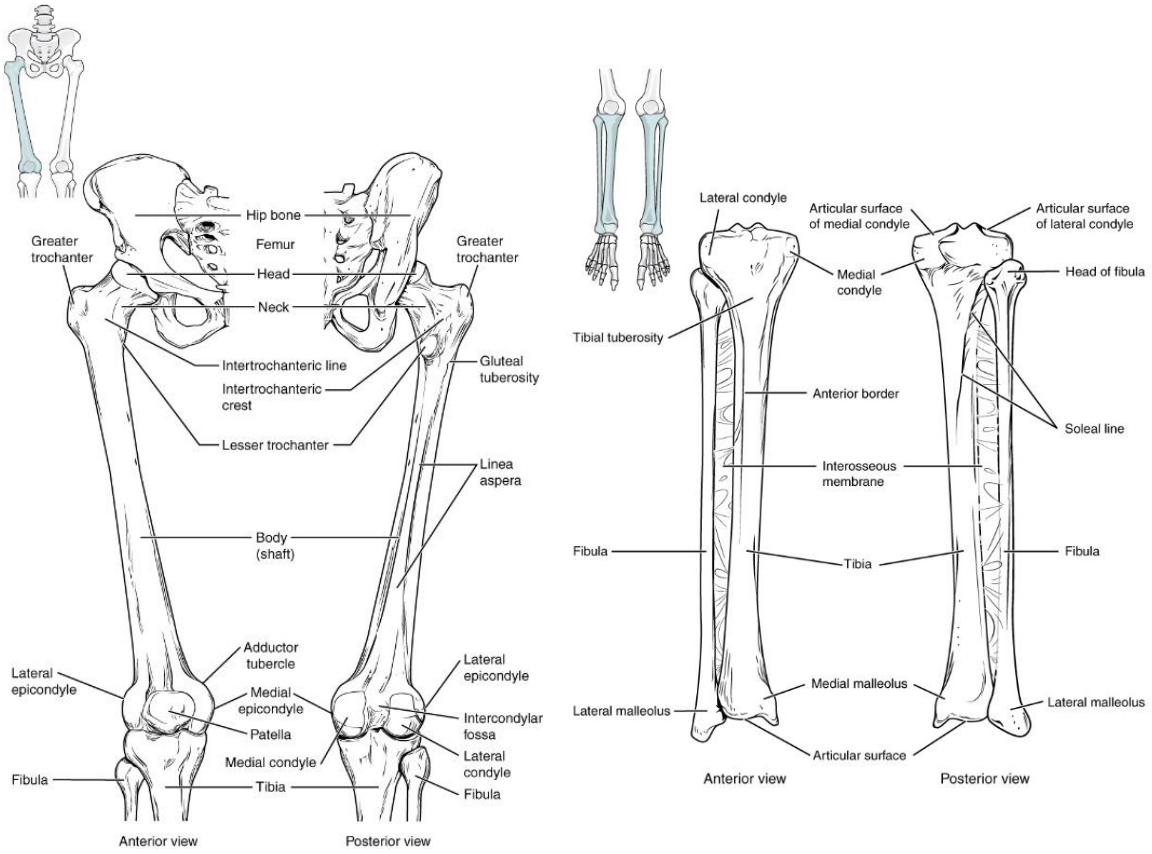


Figure 1.1. Anterior and posterior view of the right human femur (left) and tibia (right) [15].

1.1.1 Knee joint stabilisers: Ligaments and muscles of the knee

The knee joint is stabilised by ligaments and muscles. Knee joint stabilisers can be classed as primary (ligaments) or secondary (muscles) stabilisers [13]. Ligaments protect the joint against damage by preventing excess motion in the knee joint. The four main ligaments are the Medial Collateral Ligament (MCL), Lateral Collateral Ligament (LCL), Anterior Cruciate Ligament (ACL) and Posterior Cruciate Ligament (PCL) (Table 1.1). Whilst muscles also provide support and stability for the knee joint, their primarily function is to mobilise the knee joint [13]. Table 1.2 lists the muscles that play an important role in enabling knee joint motion.

Table 1.1. Major ligaments of the knee joint.

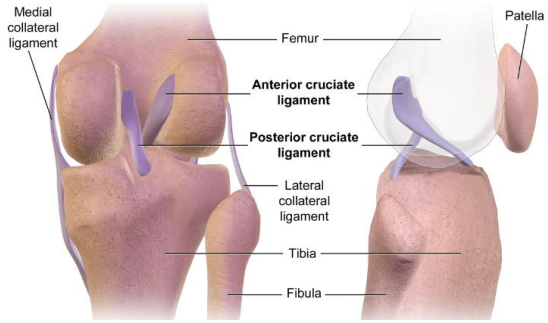
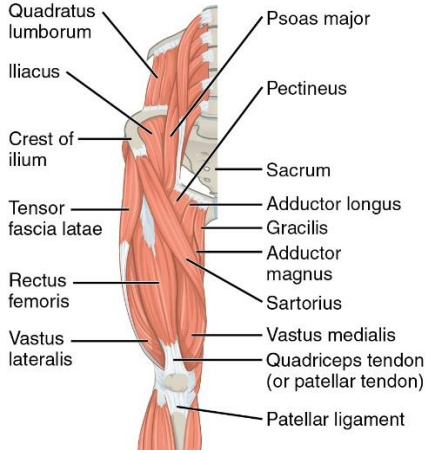
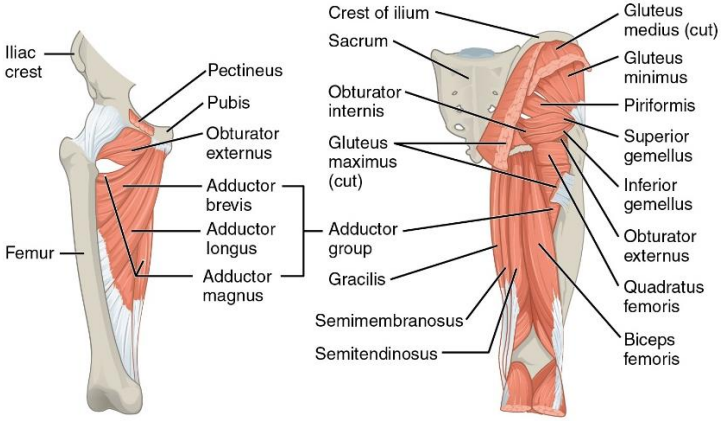
Image	Location	Ligament	Description	Attachment	Function
 ***Figure 1.2. Posterior view (left) and side view (right) of the major ligaments in the right knee.***	Extracapsular ligaments	The Medial Collateral Ligament (MCL) [15].	Tight throughout the full range of motion [17].	Medial femoral epicondyle, medial aspect of the proximal tibia [18].	Prevents excessive valgus motion of the knee joint & excessive lateral rotation [18].
		The Lateral Collateral Ligament (LCL) [15].	Slopes downwards and posteriorly [18]. Relaxed during knee flexion [17].	Lateral femoral epicondyle, head of the fibula [18].	Prevents excessive varus motion in the knee joint & excessive medial rotation [18].
	Intracapsular ligaments	The Anterior Cruciate Ligament (ACL)	Tight in full knee extension [15, 18].	Intercondylar eminence (anterior aspect), medial aspect of lateral femoral condyle [15, 18].	Prevent excessive anterolateral translation of the tibia relative to the femur [15] & knee hyperextension [15, 18].
		The Posterior Cruciate Ligament (PCL)	Tight during knee flexion [15, 18].	Intercondylar eminence (posterior aspect), medial aspect of medial femoral condyle [15, 18]. Passes medial to the ACL [18].	Prevent excessive posterior translation of the tibia on the femur [15] & knee hyper-flexion [15, 18].

Table 1.2. Muscles that control knee movement [15].

Muscle group	Location	Image	Muscle	Bony anchors w.r.t. the knee	Function
Quadriceps femoris	Anterior thigh	 ***Figure 1.3. Muscles of the anterior right thigh.***	*Vastus medialis	Patella, tibial tuberosity	Knee extension
			*Vastus intermedius	Patella, tibial tuberosity	
			*Vastus lateralis	Patella, tibial tuberosity	
			*Rectus femoris	Patella, tibial tuberosity	
Hamstring	Posterior thigh	 ***Figure 1.4. Muscles of the posterior right thigh.***	Biceps femoris	Head of fibula, lateral tibia condyle	Knee flexion [18]
			Semitendinosus	Proximal tibial shaft	Knee flexion, medial tibial rotation [18]
			Semimembranosus	Medial tibial condyle, lateral femoral condyle	Knee flexion
Adductors			Adductor longus	N/A	Knee flexion
			Adductor brevis	N/A	
			Adductor magnus	N/A	
			Pectineus	N/A	

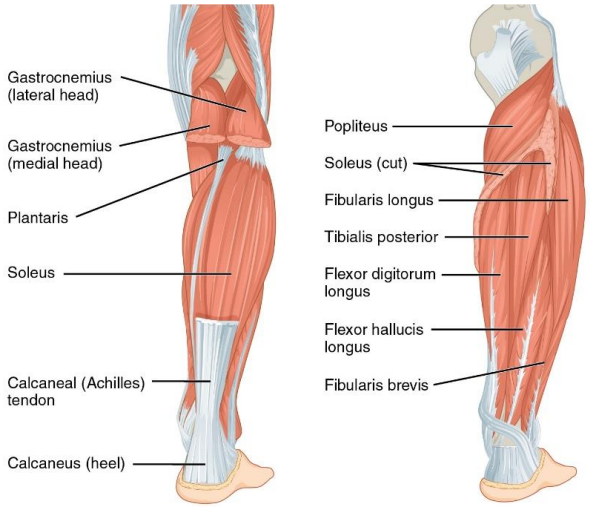
<table>
<tr><td rowspan="6">Unclassified</td><td>Anterior thigh</td><td rowspan="2">See Figure 1.3.</td><td>Sartorius [13]</td><td>Medial aspect of proximal tibia</td><td>Knee flexion [13]</td></tr>
<tr><td>Medial thigh</td><td>Gracilis</td><td>Medial aspect of proximal tibia</td><td>Knee flexion [18]</td></tr>
<tr><td rowspan="4">Posterior lower leg</td><td rowspan="4">

Figure 1.5. Muscles of the posterior right lower leg.</td><td>**Popliteus</td><td>Posterior aspect of the lateral femoral condyle</td><td>Knee flexion (initiate) [18])</td></tr>
<tr><td>Gastrocnemius</td><td>Posterior aspect of the medial and lateral femoral condyle</td><td>Knee flexion</td></tr>
<tr><td>Plantaris</td><td>Posterior femur, above lateral condyle</td><td>Knee flexion</td></tr>
<tr><td>Soleus [13]</td><td>N/A</td><td>Anterior knee translation [18]</td></tr>
<tr><td rowspan="2">Gluteal</td><td>Posterior pelvis</td><td>See Figure 1.4.</td><td>#Gluteus maximus</td><td>N/A</td><td>Knee extension</td></tr>
<tr><td>Anterior thigh</td><td>See Figure 1.3.</td><td>#Tensor fascia lata</td><td>N/A</td><td>Knee flexion</td></tr>
</table>

*Connected to the quadriceps tendon.
**Connected to the popliteal tendon.
#form the iliotibial tract (muscle) that inserts into the tibia.

1.1.2 The lateral knee

Broadly speaking, the knee joint is divided into the medial/inner half and the lateral/outer half. The lateral compartment of the knee consists of the lateral femoral condyle, the lateral tibial plateau, the lateral meniscus and the lateral soft tissues. The lateral knee is distinct from the medial knee with regard to the geometry and biomechanics [19, 20].

1.1.2.1 *The lateral femoral and tibial condyles*

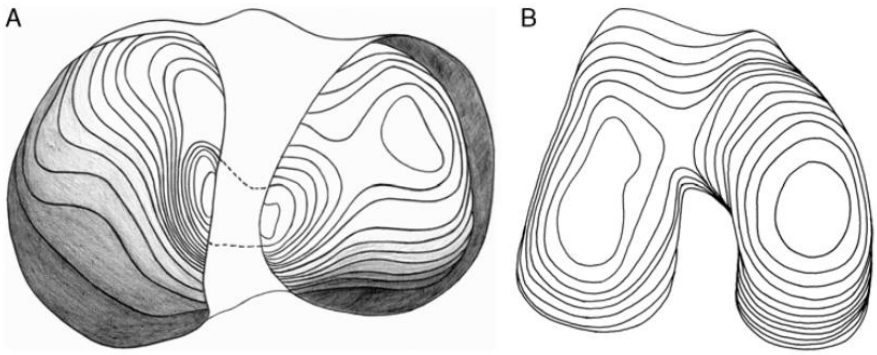


Figure 1.6. Topographical presentation of the articular surface of the right leg showing proximal tibial condyles (A) and the distal femoral condyles (B) [14]. In image A the medial condyle is on the left, and in image B the medial condyle is on the right.

Compared to the medial femoral condyle, the surface of the lateral femoral condyle is slightly smaller [15], less curved [18] (see B in Figure 1.6), less elongated [14] and less oval in shape [14]. The width of the femoral condyle is related to a subject's height and weight [21].

While the articular surface of the medial tibial condyle is concave [17], the lateral tibial condyle is slightly smaller [14] and convex in the sagittal plane [14, 17].

1.1.2.2 *The lateral meniscus*

Between the articular surfaces of the femur and the tibia, are two semi-lunar or crescent shaped fibro-cartilaginous discs called the medial and lateral menisci [22]. The menisci have many functions but their primary functions include transfer of axial loads and facilitation of load distribution in the knee joint [23]. The menisci increase the load-bearing contact area between the distal femur and the proximal tibia, which reduces the contact stress [23]. Axial compressive loading results in axial,

circumferential and radial stresses in the menisci, causing compression and circumferential expansion [24]. Secondary functions include shock absorption, stability, lubrication, nutrition [22] and proprioception [13].

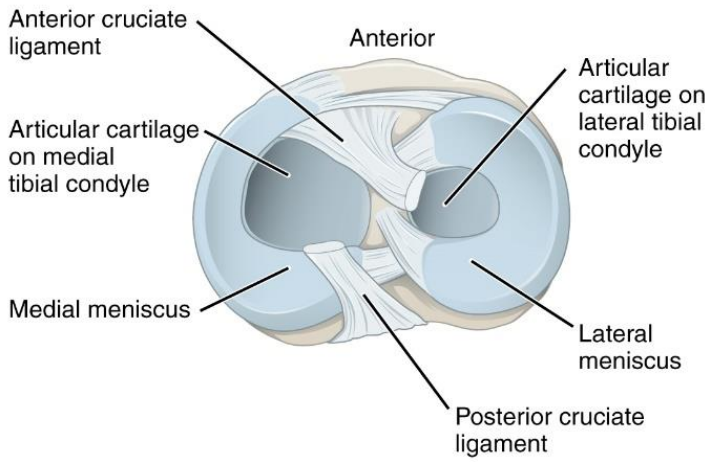


Figure 1.7. Superior view of the medial and lateral menisci on the right tibia.

The lateral meniscus is smaller than the medial meniscus (Figure 1.7) and it occupies a larger portion of the articular surface (~80%) than the medial meniscus (~60%) [14]. Due to the convex nature of the lateral tibial plateau, the lateral meniscus has more of a triangular cross section than the medial meniscus, which is more wedge shaped [14]. Sagittal section Magnetic Resonance Images (MRI) of the lateral meniscus shows a bowtie appearance [14].

The lateral meniscus has two horns, the anterior and the posterior horn. The anterior horn is attached to the tibial intercondylar eminence [14, 25]. The posterior horn of the lateral meniscus is less regular with its ligamentous attachment [25], extending either into the anterior menisco-femoral ligament of Humphrey and/or the posterior menisco-femoral ligament of Wrisberg to the lateral part of the medial femoral condyle (not depicted in Figure 1.7) [25].

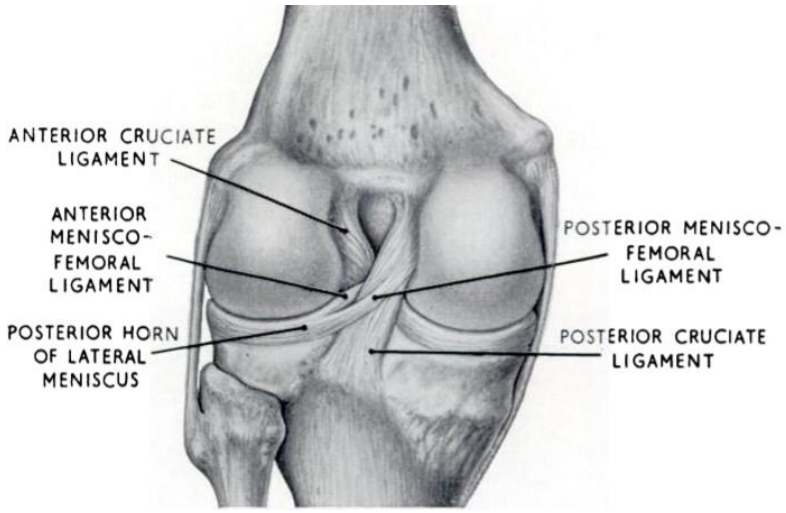


Figure 1.8. Posterior knee showing the ligamentous connections of the posterior horn of the lateral meniscus [25].

1.1.2.3 *Physiology of the lateral knee*

Dynamic motion of the knee involves the femorotibial joint and the patellofemoral joint, enabling the knee six degrees of freedom of motion [13] listed in Table 1.3.

Table 1.3. Motion in the knee joint [13].

Degrees of freedom	Plane	Motion
1	Sagittal	Knee flexion
2		Knee extension
3	Transverse	Internal rotation
4		External rotation
5	Coronal	Adduction
6		Abduction

Knee motion is traditionally described as a combination of rolling and sliding. Prior to reaching full extension, the knee is capable of limited rotation [15]. During the final portion of knee extension, the medial femoral condyle undergoes slight medial rotation relative to the tibia [15]. At full extension, the knee locks in slight hyperextension [18], to enter the most stable position of the knee joint [15]. In the locked position, the knee is completely rigid, and no rotation can occur [18].

Unlocking the knee is initiated by the popliteus muscle causing lateral rotation of the femur [18]. Thereafter, knee flexion is primarily controlled by the hamstring muscles [18]. During the first 90° of knee flexion, the tibia rotates approximately 10° internally [26]. As the knee flexes, the lateral meniscus has been shown to be more mobile than the medial meniscus: one study showed up to 20

mm posterior translation in flexion, compared with just 8 mm medially [27], while another study reported 15 mm posterior translation of the lateral femoral condyle compared with 4 mm medially [28] (Figure 1.9). The variation is likely due to the fact that the extent of meniscal movement is dependent on the loading conditions [27].

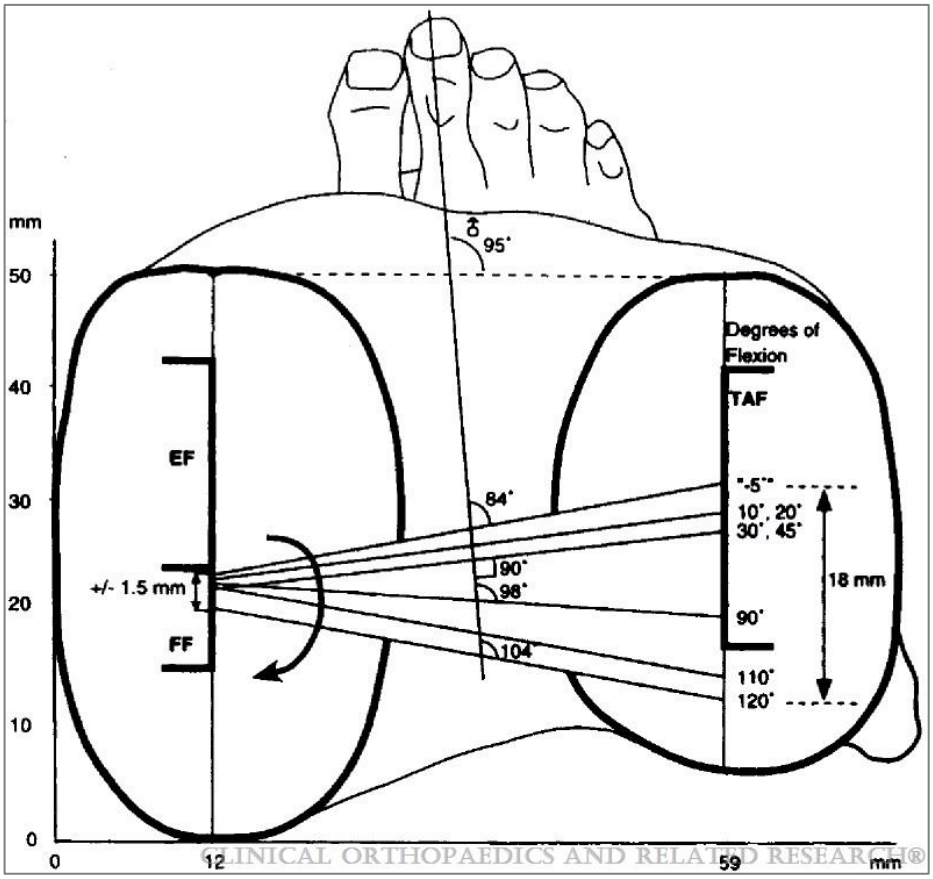


Figure 1.9. The relative movement of femoral condyles from extension to 120° flexion (right knee) [29].

To facilitate stability during dynamic motion in the knee, the medial and lateral compartments differ in the involvement of muscles and ligaments. In the medial knee, the MCL provides stability throughout the range of motion [17]. Laterally, dynamic motion is predominantly stabilised by the popliteus muscle and iliotibial tract [30]. In contrast to the medial knee, the LCL is only tight in extension, progressively becoming more lax during flexion until it reaches peak laxity at 90° of knee flexion [31]. At 90° of knee flexion, the lateral compartment of the knee can distract on average 6.7±1.9 mm (as compared to 2.1±1.1 mm medially) following application of a varus stress to a knee joint [32]. During flexion, the lateral femoral condyle translates posteriorly, until eventually in high flexion (~160° of knee flexion), it lies posterior to the lateral tibial plateau [28]. This can be seen in

Figure 1.10 where it has been found that relative to the posterior most portion of the tibial cortex, the centre of a circle outlining the lateral femoral condyle lies 7±5 mm posterior [28].

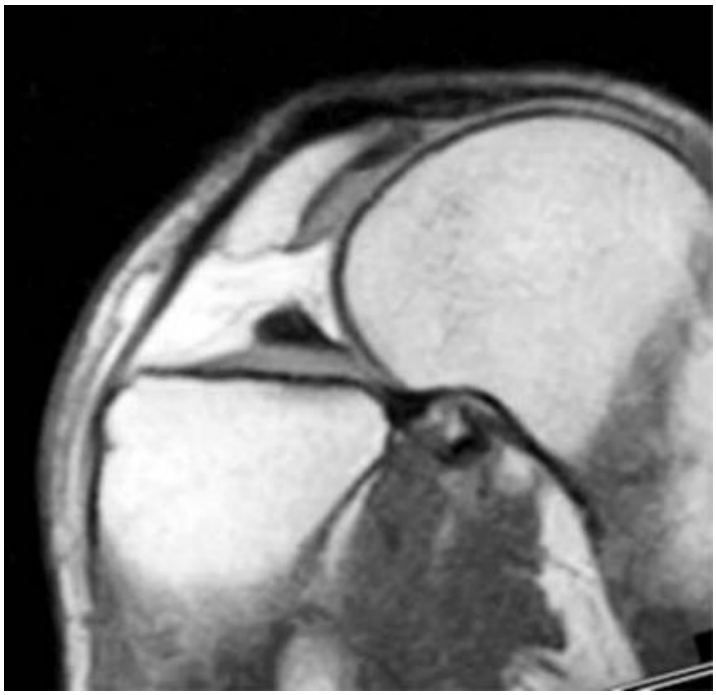


Figure 1.10. The centre of rotation of the femur lies posterior to the tibial plateau during high flexion (162° depicted) [28].

1.2 Osteoarthritis (OA) of the knee

OA is a debilitating musculoskeletal disorder. It is the most common joint disorder worldwide characterised by progressive deterioration of the articular cartilage lining the contact surfaces of bones and subchondral bone leading to bone-on-bone articulation, chronic pain, functional disability as a result of joint stiffness, crepitus and/or effusion, and ultimately joint destruction [1, 33]. The knee joint is the most commonly affected joint [2] with articular cartilage deterioration affecting either the medial compartment, lateral compartment or both compartments simultaneously.

1.2.1 Prevalence and aetiology of lateral compartment knee OA

Knee OA is highly prevalent in the ageing population and affects more women than men > 60 years of age (18% vs. 9.6%) [34]. Half of knee OA cases involve a single compartment only. Isolated lateral knee OA is less common than medial knee OA [3], accounting for 5-10% of arthritic knees

[3, 4]. The prevalence of knee OA is affected by many factors including age, sex, Body Mass Index (BMI), geography and ethnicity [2]. There also appears to be a disparity in the prevalence when comparing radiographic knee OA versus symptomatic knee OA [2]. Pooling all available data, knee OA was found to be most common in Asia, followed by Europe and finally South America (note inconclusive evidence in Oceania and Africa) [2].

Globally, there is limited epidemiological data of knee OA, which compromises our understanding of the aetiology of knee OA [2]. Genetic, biological and biomechanical components contribute to the complex epidemiology and the aetiological factors vary depending on the joint [35]. In the literature, the aetiology for lateral compartment knee OA reported can be broadly categorised as either primary [6, 36-42] e.g. rheumatoid arthritis [43, 44], osteonecrosis [6, 38, 44-46], avascular necrosis of the lateral femoral condyle [41, 42] or secondary [8, 36, 38, 39, 44-48] (following traumatic injury e.g. tibial plateau fracture, valgus deformity or a meniscectomy [41, 49]). Given the paucity of information specific to the lateral compartment of the knee, the physical appearance of lateral compartment disease may give rise to possible clues regarding its aetiology. In a study by Scott et al. in 2005, it was reported that compared to medial compartment osteoarthritis, the wear pattern in lateral knee OA, is frequently more posterior on the femoral condyle [50]. These finding were confirmed in 2009 by Gulati et al. in a study analysing the full thickness cartilage lesions in 20 knees with unicompartmental lateral knee OA, showing that the femoral lesion centre in the lateral compartment was on average located at 40° of knee flexion (SD 3°), compared to 11° (SD 3°) of knee flexion in medial knee OA. The posterior location of the femoral lesion suggests that lateral osteoarthritis is a disease of flexion [51]. Further analysis revealed that femoral lesions extended from 20° to 60° of knee flexion, which correlates to activities involving higher flexion of the knee such as stair climbing, stepping up, or getting up from a seated position [51]. Furthermore, a study conducted in Chinese patients found an association between squatting for long periods of time and an increased prevalence of radiographical knee OA [52].

1.2.2 Diagnosis

A pre-operative assessment is required to ascertain the complete patient history including gathering information on symptoms such as pain (severity, length of symptoms and location) and previous injuries sustained or surgeries [53]. Diagnosis is based on clinical symptoms and/or radiographic assessment of patients. Usually, lateral knee OA is more well tolerated than medial knee OA for a longer period of time and is often diagnosed late when the disease is more advanced [36]. While diagnosis of knee OA based on symptoms alone is possible, at this point, the disease is likely advanced and irreversible [35]. Since treatment of lateral knee OA is more likely to be successful at the early stages of disease, radiographic diagnosis is preferred if possible.

For radiographic assessment, standing weight bearing anterior/posterior (AP) radiographs often underestimate the disease severity, therefore, stress views with the knee at 40° of knee flexion or Rosenberg views are recommended. Using the radiographs, the disease severity is graded using either the Ahlbäck criteria, which grades joint space narrowing [54] or the Kellgren and Lawrence scale [55], which measures osteophyte formation and/or joint space narrowing [56].

1.2.3 Treatment

As there is no cure for osteoarthritis, the treatment is focused on reducing or eliminating pain and restoring knee functionality. The choice of treatment of lateral knee OA depends on a number of factors including disease severity [57, 58], patient age [57] and the specific clinicians' knowledge, experience and opinion [57]. Treatments usually begin with conservative management prior to more interventional treatments e.g. surgery for more advanced disease.

1.2.4 Conservative treatment/non-operative

Conservative treatments include non-pharmacological treatments, such as weight reduction, physical therapy (muscle strengthening), exercise (land-based, aerobic, water-based), walking aids, knee bracing, footwear and insoles, thermal treatments, physical rehabilitation e.g. transcutaneous

electrical nerve stimulation, acupuncture, education and self-management, and pharmacological treatments [33, 59].

In a meta-analysis assessing the use of shoe in-soles to treat medial knee OA, it was found that lateral wedges were not clinically effective at treating pain [60]. Unloading knee braces share the same functionality as wedges, which is to offload the affected knee. One study found that using unloading knee braces improved pain and functional ability [61], however, other studies found that unloading the affected knee did not prevent disease progression [62] and is less effective in obese patients [63]. While knee bracing may delay the need for surgery [64], evidence for its clinical efficacy in treating pain is not strong.

Education and self-management are central to the management of knee OA. Exercise, particularly land-based exercise, has been shown to be an effective and critical component in the conservative management of knee OA irrespective of dietary management [65, 66]. Topical non-steroidal anti-inflammatory drugs (NSAIDs) were also highly recommended [66]. Intra-articular therapies such as corticosteroid injections were recommended by the National Institute for Health and Care Excellence (NICE) in addition to core treatments [67]. The Osteoarthritis Research Society (ORS) international guidelines on OA management [66] reports that additional treatment recommendations were conditional on either patient comorbidity status or treatment time e.g. intra-articular hyaluronic acid injections are conditionally recommended for longer term treatment as they are regarded as relatively safe and evidence demonstrates improvement of symptoms beyond twelve weeks. On the contrary, oral and transdermal opioids were strongly not recommended due to high risk of chemical dependency. Finally, intra-articular stem cell therapy and intra-articular platelet rich plasma were also not recommended due to low quality evidence and lack of standardisation of the formulations.

Overall, based on results from a systematic review of guidelines [68] and more recently a comprehensive review of randomised clinical trials, systematic reviews and guidelines [33], the optimal treatment yielding the best outcomes following the conservative management of knee OA was found to involve a combination of pharmacological and non-pharmacological treatments.

1.2.5 Surgery

Over time, knee OA progressively worsens to end stage disease when patients have exhausted the conservative options available and therefore require more rigorous treatment. Surgical options include interposition arthroplasty, arthroscopic debridement [38, 57], and arthroscopic washout. Interposition arthroplasty was a technique previously described which altered the joint morphology by insertion of tissue (autograft or allograft) inside the knee joint, however, this treatment is no longer used [69]. Arthroscopic debridement and arthroscopic washout are no longer recommended (unless the patient has a clear history of mechanical locking [67]) due to high quality studies showing poor evidence of clinical efficacy when compared with optimised physical therapy and medical therapy [70]. Current treatments offered include osteotomy – high tibial osteotomy (HTO) or Distal Femoral Osteotomy (DFO), or most commonly, knee replacement surgery – either Unicompartmental Knee Replacement (UKR) or Total Knee Replacement (TKR).

HTO or DFO may be performed to re-align the mechanical axis of the lower limb and have the benefit of retaining the native knee joint structures. Some studies have reported osteotomies as an effective strategy to reduce knee osteoarthritis pain particularly in younger and more active patients [71-73], delaying the need for knee replacement surgery [74]. DFO has been reported to be effective in the treatment of lateral knee OA with associated valgus deformity [75], however, studies have also found that compared with HTO, mobile bearing UKR had superior short term clinical outcomes for advanced unicompartmental OA of the medial knee [76]. Song et al. found similar long term survival of HTO compared with UKR to 12 years, however, beyond 12 years, UKR was found to have superior post-operative survival [77].

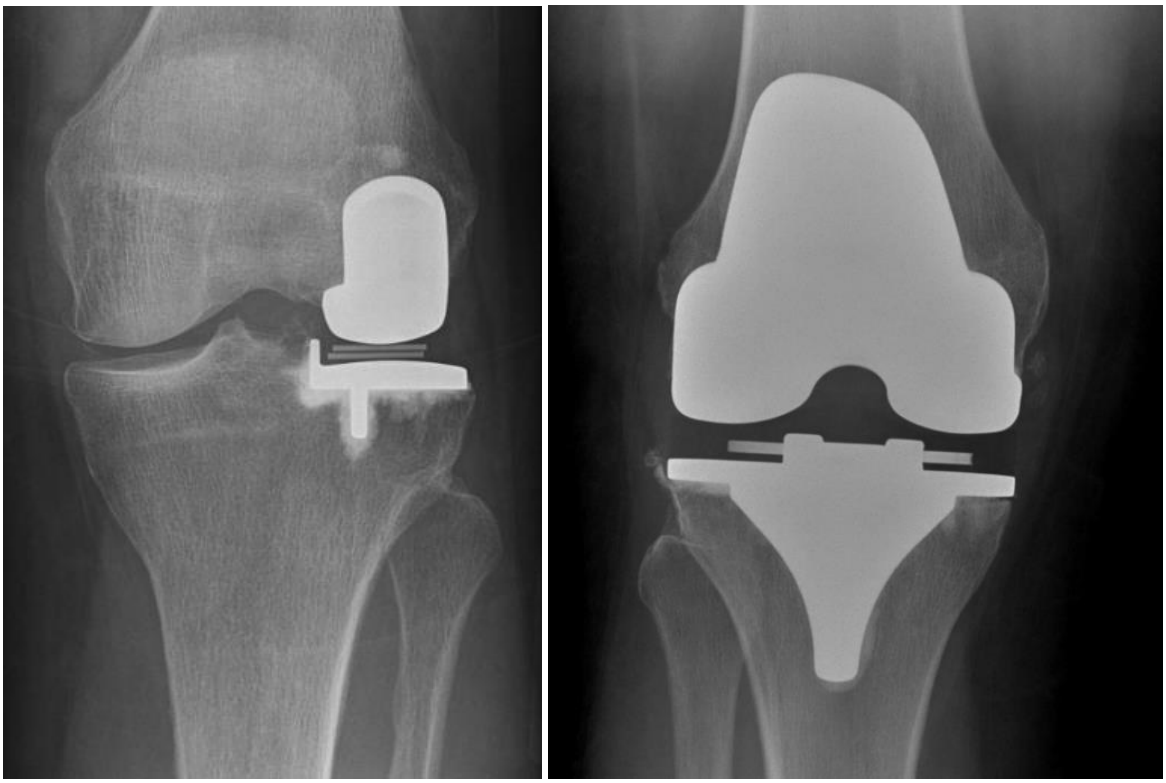


Figure 1.11. UKR (left) vs. TKR (right) X-ray.

While lateral UKR and TKR (Figure 1.11) are both reliable treatment options for end-stage osteoarthritis patients, TKR surgery has been shown to cause shortening of the patella tendon length, which is associated with subsequent joint stiffness and eventually onset of patellofemoral arthrosis [78]. Therefore, UKR, which is ligament and bone conserving, may be the better choice in certain patient groups. As osteoarthritis may affect the medial, lateral or both compartments of the knee, UKR offers the option to replace only the damaged compartment(s) of the knee. For patients with unicompartmental disease, a retrospective cohort study, comparing patients with lateral unicompartmental knee osteoarthritis receiving UKR or TKR, showed that UKR produced superior short-term functional outcomes compared with TKR, particularly in younger and in female patients [79]. Furthermore, compared with TKR, UKR patients have more natural knee kinematics with superior function [39] namely better proprioception and improved range of motion [80], improved pain relief that is more predictable [80, 81], shorter operative time, faster recovery with quicker return to daily activities [81] and it has been found to be more cost effective [82]. A propensity score-matched cohort analysis comparing UKR to TKR found UKR to have better long term outcomes and

better long term quality of life [83]. These findings were further reinforced in a systematic review comparing UKR and TKR which found that UKR consistently outperforms TKR with better functional outcomes, shorter hospital stays, lower mortality and lower early reoperation for any reason [84]. This thesis will focus on UKR and more specifically, on lateral UKR.

1.3 Lateral unicompartmental knee replacement

Due to the lower prevalence of lateral OA compared with medial UKR [3], there is relatively less information published on Lateral Unicompartmental Knee Replacement (LUKR) [57]. Given the higher prevalence of medial knee OA, UKR procedures were initially developed for the medial knee and later adapted for the lateral knee. Early results of LUKR were often mixed-cohorts including both medial and lateral UKRs. Some studies showed promising potential for LUKR [85-88] while others reported greater success with medial UKR [89, 90]. The disparity in results is likely due to the initial lack of appreciation for the unique anatomy and biomechanics of the lateral knee, which is now known to make LUKR more technically challenging than medial UKR, particularly in appropriately balancing the ligaments in the knee to achieve the correct tension [91]. However with time and greater experience, more publications focused on the lateral knee showing improvement of LUKR results [90]. Further, developments in lateral specific implant and instrumentation design alongside better patient selection has led to recent reports reflecting excellent patient outcomes, improved satisfaction and excellent early implant survivorship [6, 40-42, 92-94]. In addition to correct patient selection [57], surgeon experience [95, 96] is also essential for improved patient outcomes with some studies demonstrating a longer learning curve for less experienced surgeons/ low volume surgeons [57] and better outcomes reported for high volume surgeons [39]. Today, LUKR is an established and effective treatment for end-stage lateral knee osteoarthritis.

Many scoring methods have been developed to assess knee replacement surgery outcomes. Table 1.4 summarises the knee scores that have been reported in LUKR functional outcomes, which may facilitate understanding the results of LUKR clinical studies (including the reported functional outcomes and implant survival) summarised in Table 1.5.

Table 1.4. Knee scores used to measure LUKR functional outcomes

Knee score	Scale	Measures
Japanese Orthopaedic Association Knee Score (JOAKS) [43]	0 – 100	Pain, function
Bristol Knee Society (BKS) Score [97]	0 - 100	Pain, general and knee function
Hospital for Special Surgery (HSS) Score [43]	85 – 100: excellent 70 – 84: good 60 – 69: fair <60 poor	Pain, function, range of motion, muscle strength, knee flexion and stability.
Western Ontario and McMaster Universities Arthritis Index (WOMAC™) [98]	12-60	Function, pain and stiffness
Knee Injury and Osteoarthritis Outcome Score (KOOS) [98]	0 - 100	Based on WOMAC, but adapted for younger and more active patients. Assesses short- and long-term consequences of knee injury
Knee Injury and Osteoarthritis Outcome Score, Junior (KOOS JR) [99]	0 - 100	Shortened version of KOOS
American Knee Society Score (AKSS) [37]	0 - 100 for knee score 0 - 100 for function score Total range 0 - 200	Knee (AKSS-KS) and function (AKSS-FS): pain, stability, ROM, function
Knee Society Score (KSS) [100]	Objective knee score: 40 Satisfaction score: 40 Expectation score: 15 Functional activity score 100	Knee satisfaction and function
Oxford Knee Score (OKS) [39]	0 - 48	Knee and function
International Knee Society Score (IKSS) [101]	0 - 100	Knee and function
Short Form/SF-36 [102]	0 - 100	Quality of Life (QOL) Physical Component Summary (PCS) and Mental Health Component Summary (MCS)
Lequesne knee score [103]	0 - 24	Pain and activity
Visual Analogue Scale (VAS) for pain [42]	0 - 10	Pain intensity
Tegner activity score [104]	0 - 10	Level of activity
University of California, Los Angeles score [105]	10 options describing activity level.	Level of activity

Table 1.5. LUKR clinical outcomes and implant survival rates.

Author	Year	Implant	Bearing type	Study start date	Study end date	Number of knees	Functional outcomes	Implant survival
Gunther [19]	1996	Flat lateral Oxford (Phase I and II)	Mobile	Apr-83	Dec-91	53	In activity pain: 71% no, 24% mild, 0% moderate, 5% severe. Pain at rest: 79% none, 17% mild, 0% moderate, 5% severe.	(A) All re-operations: 5 years: 82%±15.4% (95%CI) (B) All aseptic re-operations (infection excluded): 5 years: 86%±14.1% (95%CI) (C) Re-operations for aseptic loosening: 5 years: 98%±6.7% (95%CI)
Ohdera [43]	2001	Marmor	Both	Aug-77	Dec-99	5	JOAKS: 64.2 ± 13.2 points --> 85.3 ± 9.9 points	NS
		Oxford				1		
		PCA				11		
		Omnifit				21		
Ashraf [97]	2002	St Georg Sled prosthesis	Fixed	Jan-78	Dec-99	88	BKS: 88 @ pre-op, 55 @ 2 years, 44 @ 5 years and 23 @ 10 years.	Revision for any cause - cumulative survival rate: 83% at 10 years (excl. pain), with 74.5% knees still at risk at 15 years.
Pennington [106]	2006	Miller Galante	Fixed	Aug-88	Jan-01	29	HSS: average pre-op 60 (42-79), post-op 93 (82-100) excellent (100-85 points), good (84-70 points), fair (69-60 points), or poor (60 points).	100

Forster [91]	2007	Preservation UKA	Fixed	May-01	May-03	16	AKSS knee: pre-op 41, 95 1 at year, 95 at 2 years AKSS function: pre-op 50, 100 at 1 year, 95 at 2 years OKS: pre-op 40, 16 at 1 year, 15 at 2 years	NS
			Mobile			12	AKSS knee: pre-op 48, 95 at 1 year, 95 at 2 years AKSS function: pre-op 60, 100 at 1 year, 100 at 2 years OKS: pre-op 41, 19 at 1 year, 13 at 2 years	NS
Sah [107]	2007	Brigham unicondylar prosthesis	Fixed	NS	1981	49	AKSS knee: pre-op 39, 89 post-op AKSS function: pre-op 45, post-op 80	NS
		Press Fit Condylar (PFC) prosthesis	NS	1990s				
		PFC Sigma Prosthesis	NS	2001				
		Preservation Unicompartmental knee	Fixed or mobile	2002	2004			
Volpi [57]	2007	Miller Galante	Fixed	Mar-00	Feb-05	159	HSS: average pre-op 52.9 (48 - 68), post-op 88.04 (71-95)	NS
Argenson [45]	2008	Marmor-like	Fixed	Feb-82	Dec-04	15	KSS-KS: pre-op 57±10 (35-75 SD), post-op 88±5 (40-100 SD) KSS-FS: pre-op 46±5 (10-89 SD), post-op 78±3 (20-100 SD)	Revision for any reason: 92% at 10 years (95%CI: 0.79-0.99), 84% at 16 years (95%CI: 0.71-0.97)
		Alpina	Fixed			1		
		Miller Galante	Fixed			20		
		ZUK	Fixed			4		
Pandit [7]	2010	Flat lateral Oxford Phase III	Mobile	Apr-98	Oct-04	65	86%: no pain/mild pain at rest and on activity. Mean flexion improved from 111° pre-op to 117° post-op. OKS improved from 24 - 36.	NS
John [108]	2011	Miller Galante	Fixed	1989	2006	94	Average total BKS at follow-up was 43.61 (range: 28–50). The mean functional score was 16.3 (9–20). Patient satisfaction: 88% of patients had good or excellent total score and 73% patients had good or excellent functional score. Fifteen patients had a fair/poor functional score while only eight had their total score in the fair/poor. The mean range of movement at follow-up was 110.6° (80–130°). Overall, 25 (45%) of the knees had no pain, 24 (43%) mild or occasional pain and 7 (12%) moderate pain. All patients stated that they had improved after the procedure.	97% at five years, 41% at eight years

Lustig [109]	2011	Cemented all-polyethylene tibial component	Fixed	Jan-90	Oct-03	54	IKSS knee average: 68.2 (range 35-92) pre-op to 94.9 points (range 70-100 points) Mean range of flexion of 132.6° (range, 115°-150°) Mean IKSS function score: 68.9 (range 0-100) pre-op to 81.8 points (range 25-100 points) post-op.	Prosthesis removal endpoint: 98.08% (94.38%-100%) at 10 years
Panni [110]	2011	ZUK metal backed tibial tray	Fixed	Feb-05	Dec-07	80	IKSS Knee 46 pre-op to 80 post-op. IKSS Function 54 pre-op to 94 post-op.	100% at last follow up
Berend [58]	2012	Repicci II	Fixed	Apr-04	Dec-08	42	Average KSS: - Pain: from 8 (range, 0–45) pre-op to 46 (range, 10–50) post-op for pain - Total clinical: 49 (range, 10–95) pre-op to 94 (range, 10–95) post-op - Function: 47 (range, 0–90) pre-op to 89 (range, 40–100) post-op Average ROM: 115° (range, 70°–125°) pre-op to 124° (range, 95°–45°) post-op	NS
		Vanguard M ™	Fixed			90		
Heyse [95]	2012	Genesis Unicondylar implant/Accuris System - full poly	Fixed	1993	2004	223	Average KS-KS: - All-poly: 97.0±2.4 male, 96.8±2.9 female Average KS-FS: - All-poly: 98.6±2.5 male, 96.5±4.2 female	Overall LUKR: 91.8% at 10 and 15 years. All poly: 93.3% at 10 and 15 years
		Genesis Unicondylar implant/Accuris System - metal-backed cemented					Average KS-KS: Metal backed and cemented: 93.5±0.7 male, 95.5±4.8 female Average KS-FS: Metal backed and cemented: 95.0±0.0 male, 91.2±7.5 female	Metal backed cemented: 88.7% at 10 years and 82.4% at 15 years
		Genesis Unicondylar implant/Accuris System - metal-backed uncemented					Average KS-KS: - Metal backed and uncemented: 93.8±2.8 male, 90.4±3.6 female Average Knee Society function scores: - Metal backed and uncemented: 98.8±2.5 male, 93.6±3.8 female	Metal backed uncemented: 97.3% at 10 years and 74.9& at 15 years
Lustig [48]	2012	HLS Uni Evolution	Fixed	Jan - 85	Dec-07	13	Mean IKSS knee scores improved from 51 points (range, 29–75 points) pre-op to 88 points (range, 65–100 points) post-op Mean IKSS knee function scores improved from 51 points (range, 10–89 points) pre-op to 87 points (range, 35–100 points) post-op	Revision for any reason as the end point, the implant survivorship rates were 100% at 5 and 10 years and 80% at 15 years
		Marmor II						
		Miller-Galante unicompartmental knee system						

Liebs [44]	2013	Preservation (R) mobile bearing prosthesis	Mobile	Apr-02	Oct-09	558	WOMAC™ function score (points): 33.6 ± 29.5 WOMAC™ pain score (points): 33.7 ± 31.3 WOMAC™ stiffness score (points): 22.8 ± 27.1 SF-36 PCS (points): 38.0 ± 11.4 SF-36 MCS (points): 48.1 ± 12.2 Lequesne knee score (points): 7.9 ± 4.1	Implant survival was 91.8% at 5 years and 83% at 9 years.
Schelfaut [8]	2013	Domed lateral Oxford UKR	Mobile	Jan-09	Apr-11	25	General patient satisfaction at the latest follow-up was excellent in 84 %, good in 12 % and fair in 4%. The OKS improved significantly ($p<0.0001$) from a pre-op mean of 23.3 (range 8–40, SD 8.4) to 42.1 (range 23–48, SD 6.7) post-op. The mean post-op flexion was 127° (range 110°–155°, SD 10.5°).	NS
Lustig [111]	2014	HLS Uni Evolution	Fixed	Jan-88	Oct-03	54	Mean KSS knee score was 95.1 points and mean KSS function score was 82.2 points. The mean ROM was 132.6° (range, 115–150°).	Implant survival was 94.4% at ten years and 91.4% at 15 years.
Marson [112]	2014	Domed lateral Oxford UKR	Mobile	Oct-07	Aug-11	15	Average OKS was 36.6 (95 % CI 29.0–44.2)	100% at 2 years
		Zimmer unicompartmental high flex knee system	Fixed			12	Average OKS was 28.6 (19.8–37.5) for the Zimmer fixed-bearing prosthesis ($p = 0.15$).	90% at 2 years
Smith [37]	2014	AMC Uniglide prosthesis	Fixed	Dec-02	Dec-11	101	The mean total AKSS increased from 100 (30–182) pre-op, to 159 (69–200) at 5 years ($p < 0.001$). Mean OKS was 20 (5–45) pre-op and 37 (9–48) at 5 years ($p < 0.001$). WOMAC scores were 36 (15–53) pre-op, and 22 (12–48) at 5 years ($p < 0.001$).	All 101 UKRs were included in survivorship calculations. Taking removal of prosthesis as the end-point, implant survival was 98.7% at 2 and 95.5% at 5 years.
Walker [113]	2014	Domed lateral Oxford UKR	Mobile	Dec-06	Mar-09	50 patients	Mean postoperative OKS and ROM: mean OKS was 43 [standard deviation (SD) 4] ROM was 127° (SD 13)	Revision for any reason as the endpoint was 96 % [95%confidence interval (CI) 72–99] at 2 years
Kim [38]	2016	Zimmer unicompartmental	Fixed	Jan-11	Feb-14	30	The mean KS knee score increased from 63.2 points (range, 48 to 70 points) pre-op to 86.0 points (range, 74 to 95 points) at the final follow-up ($p < 0.001$).	NS

		high-flex knee (ZUK) prosthesis					The mean KS knee function score increased from 68.6 points (range, 35 to 80 points) pre-op to 92.4 points (range, 60 to 100 points) at the final follow-up ($p < 0.001$).	
Masud [114]	2018	Domed lateral Oxford UKR	Mobile	2007	2014	27	Pre-op OKS 20, post-op OKS 37	96.30% overall implant survivorship
Newman [39]	2017	Domed lateral Oxford UKR	Mobile	2005	2009	64	Median OKS was 26 (range nine to 36) pre-op and 42 (10–48) at final follow-up.	Re-operation plots a cumulative survival of 0.80 at 119 months.
Edmiston [93]	2018	Miller-Galante Unicompartmental Knee	Fixed	2003	2004	65 patients	Mean KSS pre-op (SD) 124.8 (23.4) and 144.7 (30.8) post-op	Overall survivorship was 94% at a mean of 82 months
		Cemented Zimmer Unicompartmental Knee		2004	2014		Mean KSS pre-op was 126.6 (25.1 SD) and 155.2 (36.1 SD) post-op	
Fornell [41]	2018	Domed lateral Oxford UKR	Mobile	Jun-08	May-15	41	At latest follow-up: The VAS score mean pre-op of 9.3 (range: 8 - 10) to post-op of 1.25 (range: 0 - six). The mean pre-op OKS was 26 (20–36) and the mean post-op score was 44 (38–48).	Survival, with revision for any reason as the endpoint, was 97.5% at five years and 86.4% at 6 years.
Saragaglia [115]	2018	KAPS® prosthesis	Fixed Mobile	Mar-05	Mar-10	103	The mean post-op OKS was 18 ± 5 (13–30)	NS for lateral only.
Walker [40]	2018	Domed lateral Oxford UKR	Mobile	Nov-06	Jan-14	344	Mean OKS of 40.3 (95% CI 39.4 to 41.2), a mean Tegner activity score of 3.2 (95% CI 3.1 to 3.3) and a mean University of California, Los Angeles score of 5.7 (95% CI 5.5 to 5.9)	Survival rate of 90.1% at three years (95%CI: 86.1 to 93.1; number at risk: 155) and 85.0% at five years (95% CI 77.9 to 89.9; number at risk: 43)
Deroche [116]	2019	HLS Uni Evolution	Fixed	Jan-88	Oct-03	54	The mean functional IKSS score was 66.5 ± 26.8, with a mean objective score of 84.4 points ± 13.2.	Cumulative survival rate was 82.1% at 15 years and 79.4% at 20 years
Gill [92]	2019	Physica ZUK	Fixed	Jun-05	2017	14	KS-KS and KS-FS pre-op was 46.4 and 48.7, and 91.3 and 93.1 respectively post-op ($p = 0.0001$)	100%
Burger [94]	2020	Metal-backed tibial onlay components	NS	Jun-07	Aug-16	171	Of 152 knees: Mean post-op KOOS score (SD) 85.6 (14.3) 150/162 (92.6%) of patients were either very satisfied or satisfied with their knee function	Five-year survivorship was 97.7% (95% CI 94.9 to 100)

Burger [117]	2020	All-poly	Fixed	2007	2017	11	NS	Five-year crude cumulative revision rate using the Kaplan-Meier analysis was 12.9% (95% confidence interval [CI] 6.6-16.2)
	Metal-backed	91	NS					
		Domed lateral Oxford UKR	Mobile			435	NS	
Deroche [46]	2020	HLS Uni Evolution	Fixed	Jan-88	Sep-14	91	At last follow-up: mean IKSS knees score was 87.0 ± 10.9 (range, 45–100) and mean function score 80.2 ± 14.3 (range, 30–100). Mean maximum flexion was 125 ± 11◦ (range, 85°–140◦). 94.3% of patients were satisfied or very satisfied.	All-cause revision surgery was 85.4% at 10 years and 79.4% at 20 years
		Alpina Uni®				80		
		ZUK®				78		
		Sigma HP®				15		
		Uni Score®				4		
Kagan [118]	2020	Zimmer Natural-Knee UKA	Fixed	Jul-00	Dec-10	177	KOOS, JR was 101.3 (94.5 to 108.02). Satisfaction was 92% (28/30) in the lateral UKRs (P = 0.309).	The cumulative incidence of revision TKA at 10 years was approximately 5% for the lateral UKRs (SHR, 0.67; 95% confidence interval, 0.22-20.2, P = .476) when including death as a competing risk
Kennedy [6]	2020	Domed lateral Oxford UKR	Mobile	Sep-04	Dec-15	325	Median OKS was 43 (interquartile range (IQR) 37 to 47), with 260 (80%) achieving a good or excellent score (OKS > 34).	Ten- year survival was 85% (95% confidence interval (CI) 79 to 90, at risk 72)
Walker [42]	2020	Oxford Fixed lateral prosthesis	Fixed	Feb-13	Jun-17	52	OKS pre-op was 26.4 ± 6.9 (12–41) and 39.7 ± 8.4 (15–48) post-op AKSS-O pre-op was 54.3 ± 15.3 (18–90) and 82.2 ± 15.6 (40–100) post-op AKS-FS was 56.4 ± 21.3 (10–100) pre-op and 83.1 ± 20.2 (five to 100) post-op ROM was 123.5 ± 13.5 (90–140) pre-op and 134 ± 10.3 (95–150) post-op	Survivorship was 100% at 2 years for revision for any reason (exchange any component) as the end point. Survivorship was 94.2% (95% confidence interval (CI): 83.2–98.1%)

								with the endpoint 'any reoperation'.
Mohammad [119]	2021	Domed lateral Oxford UKR	Mobile	Jan-05	Dec-17	992	NS	The 5 and 10 year cumulative implant survival rates were 92.4% (CI: 90.3-94.1) and 88.6% (CI: 85.3-91.2), respectively
Murray [120]	2021	St Georg Sled UKA	Fixed	1974	1994	82	Mean BKS (Best 100), mean OKS (Best 48) and WOMAC (Best 12): Pre-op: 52.7, N/A, N/A 10 years: 84.5, 35.3, 23.2 15 years: 76.6, 27.3, 30.3 20 years: 74, 21, 34 25 years: 90, 37.7, 18.75	Survivorship was 72% at 15 years, and 68% at 20 and 25 years
Xue [121]	2021	St Georg Sled UKA	Fixed	Jan-13	2020	260	At last follow-up, the HSS score was 52.1 (range, 38-80) pre-op and 85.6 (range, 61-98) post-op The OKS: 22.8 (range, 16-32) pre-op to 42.7 (range, 30-47) post-op	The 5 year survival was 99.5%.

NS = Not Stated

1.3.1 LUKR Implants

Unicompartmental knee replacement implants enable targeted treatment of the damaged tissue in an osteoarthritic knee, by replacing just the damaged tissues in the knee joint. The McIntosh surface arthroplasty [122] and the McKeever implant [123] were some of the first resurfacing implants designed in the 1950s. The first modern UKR was the Marmor modular knee developed in the early 1970s [124]. Since then, many UKR implants have been developed, with most implants designed for the medial knee and later adapted for use in the lateral knee [57]. The LUKR implants used to date either have a mobile bearing, fixed bearing or semi-constrained bearing. In 1978, Goodfellow et al. suggested that to counter dislocation, bearing constraints in the form of either stops or tracks, could be added [125]. Initial experiments with semi-constrained bearings (termed "washers") noted rocking of the tibial component, therefore, the authors concluded that it would be ideal if the constraints could be activated outside the normal ligamentous range. Later, two products with semi-constrained bearings were created: the Preservation UKA (DePuy) and the New Jersey Low contact Stress Uni (DePuy International, Leeds, UK). Both implants featured a curved groove/track in the tibial component to partially restrain the bearing movement: enabling AP movement but not mediolateral movement [91, 123]. While initial results were promising [123], the Preservation Uni was recalled in 2009, due to poor implant performance and similarly, early failure of the LCS mobile bearing prosthesis have been reported with over-constraint of the mobile bearing cited as the cause [126]. Subsequent designs were either mobile or fixed bearing designs (Table 1.6 and Table 1.7, respectively), with some implants being hybrid (Table 1.8). Today, fixed bearing designs dominate the market at 85% (Table 1.6), despite mobile bearing designs having theoretical advantages.

Table 1.6. Fixed bearing LUKR implants reported in literature and/ or registries.

Implant	Manufacturer	Mediolateral	Reference	Registry
Allegretto	Sulzer Orthopedics	Medial or lateral	N/A	Switzerland [127], Australia [128], Italy [129]
Alpina	Biomet, France, Valence	Medial or lateral	[45, 46, 118, 130]	Switzerland [127]
AMC Uniglide prosthesis	Corin PLC, Cirencester, UK	Medial or lateral	[37]	Australia [128], UK [5]
Balansys Uni System	Mathys European Orthopedics	Medial or lateral	N/A	Switzerland [127], Australia [128], Italy [129]
Brigham Prosthesis	NS	Medial or lateral	[89]	N/A
Duracon UKA	Howmedica, Rutherford, NJ	Not found	[43, 131]	N/A
Endo-Model Sled	Link	Medial or lateral	N/A	Australia [128], Italy [129]
Genesis Unicondylar implant/Accuris System	Smith & Nephew, Memphis, TN	Medial or lateral	[95]	Australia [128], Italy [129], Norway [132], Sweden [133]
GKS	Permedica	Medial or lateral	N/A	Italy [129]
GMK Uni	Medacta International	Medial or lateral	N/A	Switzerland [127]
HLS Uni Evolution	Tornier, Inc, Grenoble, France	Medial or lateral	[46, 48, 116].	Portugal [134], Italy [129]
iBalance Uni	Arthrex	Medial or lateral	N/A	Switzerland [127]
Journey Uni	Smith & Nephew	Medial or lateral	N/A	Slovenia [135], Switzerland [127], USA [136], New Zealand [137], Germany [138], Australia [128], UK [5], Belgium [139], Italy [129], Norway [132], Netherlands [140]
Link Sled	Link	Medial or lateral	N/A	New Zealand [137], Germany [138], Norway [132], Sweden [133]
Marmor, Marmor II	Smith & Nephew Richards, Orthez, France	Medial or lateral	[48, 141, 142]	N/A

Miller-Galante	Zimmer, Warsaw, Indiana	Medial or lateral	[48, 106, 142]	Australia [128], UK [5], Italy [129], Norway [132], Sweden [133]
Optetrak uni	Exactec	Medial or lateral	N/A	Italy [129]
Oxford fixed bearing implant	Biomet Inc., Warsaw, IN, USA	Lateral	[42]	Germany [138]
Physica ZUK	Lima Corporate, Udine, Italy	Medial or lateral	[92]	Slovenia [135], Switzerland [127], Germany [138], UK [5], Belgium [139], Italy [129], Netherlands [140]
Porous coated anatomic UKA	Howmedica, Rutherford, NJ	Medial or lateral	[143]	N/A
Repicci II®	Biomet, Inc, Warsaw, IN	Medial or lateral	[144]	Australia [128]
Restoris MCK	Stryker	Medial or lateral	N/A	USA [136], New Zealand [137], Australia [128], Italy [129]
Sigma HP ®	DePuy Synthes	Medial or lateral	[46]	Switzerland [127], USA [136], New Zealand [137], Germany [138], Australia [128], UK [5], Italy [129], Norway [132], Sweden [132]
St Georg Sled prosthesis	Waldemar, Link, Hamburg, Germany	Medial or lateral	[90, 97]	N/A
Triathlon PKR	Stryker	Medial or lateral	N/A	Switzerland [127], USA [136], New Zealand, Germany [138], Australia [128], UK [5], Belgium [139], Norway [132], Sweden [132], Netherlands [140]
UC Plus Solution	Smith & Nephew	Medial or lateral	N/A	Italy [129]
Uni Score®	Amplitude	Medial or lateral	[46]	Slovenia [135]
Unix UKR	Stryker Orthopedics, Mahwah, NJ	Medial or lateral	[118, 145]	Australia [128]
Vanguard M™	Biomet, Inc, Warsaw, IN	Medial or lateral	[144]	N/A
Zimmer unicompartmental high flex knee system (ZUK)	Zimmer, Indiana	Medial or lateral	[38, 112]	Australia [128], Norway [132], Sweden [133]

Zimmer Natural-Knee UKA	Zimmer Biomet, Warsaw, IN	Medial or lateral	[118]	N/A
ZUK	Smith & Nephew	Medial or lateral	[45, 116]	N/A

N/A = Not Applicable

Table* 1.7*. Mobile bearing LUKR implants reported in literature and/ or registries.

Implant	Manufacturer	Mediolateral	Literature reference	Registry
AMC unicompartmental knee	Alphanorm, Quiershied, Germany	Medial or lateral	[20]	N/A
New Jersey low contact stress knee	DePuy International, Leeds, UK	Medial or lateral	[91, 123, 141]	N/A
Oxford Domed Lateral implant	Zimmer Biomet, Bridgend, UK	Lateral	[146]	*Germany [138], Australia [128], UK [5], USA [136], Italy [129], Netherlands [140], New Zealand [137], Norway [132], Portugal [134], Slovenia [135], Sweden [133], Switzerland [127]
Oxford implant	Zimmer Biomet, Bridgend, UK	Medial	[147]	

N/A = Not Applicable
* Not clear which Oxford implant (mediolateral)

Table 1.8. LUKR implants with mobile/fixed bearing options reported in literature and/ or registries.

Implant	Manufacturer	Bearing type	Mediolateral	Literature reference	Registry
KAPS® uni knee arthroplasty system	X.NOV (Héricourt, France)	Mobile or fixed	Medial or lateral	[115]	N/A
Preservation UKA	DePuy Orthopaedics Inc., USA	Fixed or Mobile	Medial or lateral	[44, 91, 148]	Portugal [134], Australia [128], Italy [129], Norway [132]

N/A = Not Applicable

1.3.2 Fixed versus mobile bearing LUKR

Some authors have argued that implant choice can have a significant impact on post-operative outcomes [91, 95], however, most studies have found little to no difference in functional outcomes [149-153], clinical and radiological outcomes [154], or complication rates [152] between fixed and mobile bearing implants. What does differ in the literature however, is the failure modes and the time to failure: Zhang et al. [154] reported different complications/ failure modes for fixed versus mobile bearing designs: fixed bearing implants fail due to progression of arthritis, polyethylene wear, aseptic loosening, whereas mobile bearing implants fail due to progression of arthritis, aseptic loosening, bearing dislocation and persistent pain. There are many studies reporting higher implant survivorship for fixed bearing designs than mobile bearing designs [42, 92, 117] particularly early on. The discrepancy is probably because mobile bearing implants tend to fail early due to bearing dislocation whereas fixed bearing implants fail later due to polyethylene wear [154]. A more recent study comparing cemented lateral unicompartmental implants found no significant difference in the implant survival of fixed and mobile bearing designs at 10 years (93.8% for fixed bearing versus 90.6% for mobile bearing) [153]. Similarly, Zhang et al. reported no significant difference in implant survivorship at 20 years of 83% for the fixed bearing implant and 80% for the mobile bearing implant [154]. It follows that since bearing dislocation is a problem unique to mobile bearing UKRs, and fixed bearing UKRs fail as a result of polyethylene wear [154], if the issue of mobile bearing dislocation can be resolved, it may allow mobile bearing implants to outperform fixed bearing implants particularly when using long term measurements such as implant survivorship. This is particularly pertinent given an ageing population and a greater demand for UKRs in younger and more active patient populations [155].

1.3.3 The Oxford Knee implant



Figure 1.12. The Oxford Knee. The medial implant features a flat tibial surface [156].

The Oxford Knee implant (Figure 1.12) is the most popular UKR device in clinical use [5, 137]. The implant consists of fully congruent components with a spherical femoral component, a mobile bearing and a tibial component. The Oxford Knee implant was the first partial knee implant to have a completely separate mobile and unconstrained bearing component, allowing the mobile bearing to slide on the tibial component surface without artificial constraints on the bearing [90]. Furthermore, this allowed the ligaments to determine the axis of rotation for bone movement [90]. This unconstrained motion distinguished the Oxford Knee by restoring nearly normal knee motion [141] and reduce bearing wear rate. Initially designed for use bicompartmentally, the implant was the same when used medially or laterally. Intraoperative transverse plane assessment of the mobile bearing in a dynamic knee has found that the bearing moves posteriorly (in both the medial and in the lateral knee) with increasing angles of knee flexion [157, 158]. Medially, this posterior movement of the mobile bearing has been found to occur usually with the bearing contacting the tibial wall [157], however, this movement has not been studied laterally. Further, the vertical position of the mobile bearing in a flexed knee has not been characterised in either the medial or the lateral knee. Implanted

medially, the device performed well [122], however the dislocation rate when used laterally was high [19]. In June 1976, the first device was successfully implanted into a patient [90].

1.4 The Oxford Domed Lateral (ODL) implant



Figure 1.13. The Oxford Domed Lateral implant [113].

Following ongoing dislocations of the Oxford knee implant when used in the lateral knee, in 2006, the ODL UKR implant (Figure 1.13) was introduced. The characteristic design of the implant featured a spherically convex domed tibial surface [159]. To maintain full congruity with the spherical surfaces of the tibial and femoral components, both articular surfaces of the bearing were made spherically concave. It was hoped that this modification would not only more accurately restore the normal convex tibial surface to improve kinematics, particularly in high flexion, but most importantly, that it would reduce the high medial dislocation rate by increasing the "entrapment" of the mobile bearing (the maximal height of the mobile bearing). By increasing the bearing entrapment, this would make dislocations more difficult, thereby decreasing the dislocation rate clinically [159].

1.4.1 ODL implant: Surgical indications and the procedure

The indications and contra-indications for surgery have been widely debated however, a recent systematic review by Buzin et al. [53] cited the following as indications for surgery: mostly isolated lateral compartment disease, <10° angular deformity and <80kg body weight. Contraindications were noted as >15° flexion contracture, ACL and/or MCL insufficiency and inflammatory synovitis. However, the official indications as listed by the designer surgeons are based on the extensive knowledge of the medial knee because relatively low numbers of lateral UKRs are done. The indications are: bone on bone lateral knee osteoarthritis, full thickness cartilage in the medial knee, correctable intra-articular deformity, intact ACL with no contraindications. Medially, Spontaneous Osteonecrosis of the Knee is also listed as a good but rare indication, however, this is not specifically listed as an indication laterally, probably due to the low numbers [17].

The full surgical technique [17] for the ODL implant involves placing the knee in a leg holder which enables the knee to hang in 100° - 110° flexion. A lateral parapatellar skin incision is made, lateral osteophytes removed and ACL status checked. The vertical tibial cut is made through the centre of the patellar tendon and an internally rotated and the horizontal tibial cut is made with 7° posterior slope. The knee is brought into full extension to check for ligament tension, and the appropriate bearing thickness is chosen with the knee in extension. The femoral component is anatomically placed. Both the femoral and tibial components are fixed with polymethylmethacrylate (PMMA) bone cement.

1.4.2 ODL implant complications: Mobile bearing dislocation

In addition to changing the design of the implant to counter dislocations, many attempts have been made to address the problem of mobile bearing dislocation. Changes to the surgical technique include introduction of a lateral parapatellar approach [19, 122], internally rotating the tibial cut [7] and avoiding excessive elevation of the lateral tibial joint line, which has been shown to increase the risk of dislocation [96, 160]. Prior to the introduction of the lateral approach, the popliteus tendon was

also routinely divided to avoid tendon bowstring action [19], however, this is no longer done. Surgeons have also attempted inserting screws into the tibial eminence, with the screw heads positioned above the tibial wall to augment the tibial wall height, to prevent medial dislocations (Figure 1.14) [7]. Some have reported success with this procedure [7, 8], however, some concern around potential metallosis resulting from metal-on-metal collision have also been expressed [7].



Figure 1.14. Screws inserted into the tibial eminence to discourage medial dislocation of the mobile bearing [8].

In addition, the instrumentation was improved through the introduction of the "Microplasty instrumentation" [17, 19] and surgeon experience has also increased over time [161]. Most recent

studies report a similar dislocation rate of between 2 - 4% [6-8, 39, 41, 119]. Currently, the ongoing dislocation of the mobile bearing in the ODL implant has led to introduction of the Fixed Lateral Oxford (FLO) UKR, which is intended for use interchangeably with the mobile ODL implant. If the bearing is deemed unstable intra-operatively, surgeons are at present recommended to use the FLO interchangeably with the ODL implant [6].

1.4.2.1 *Dislocation direction*

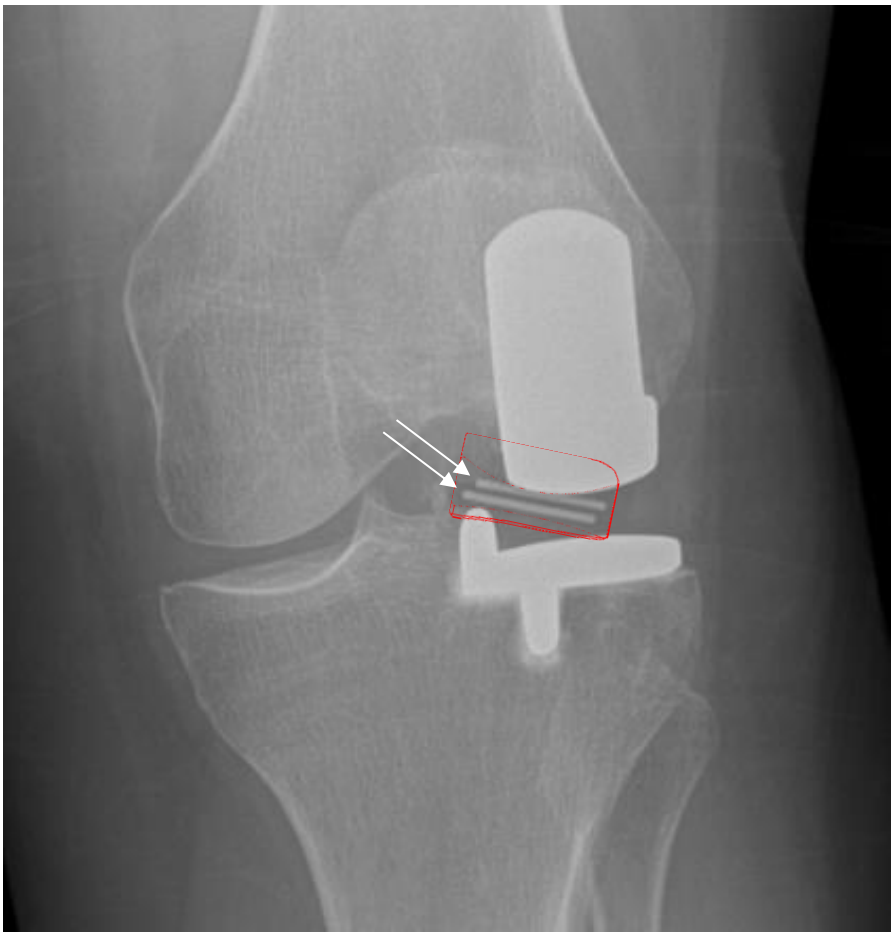


Figure 1.15. Annotated X-Ray of a patients left knee. Arrows indicate the radio-opaque markers and the dislocated bearing is outlined in red.

Most ODL bearings dislocate medially - often described as the bearing dislocating into the intercondylar notch [6-8, 41] whereby the bearing becomes lodged medially above the tibial

component wall at an angle (Figure 1.15). Clinically, bearings do not dislocate laterally probably due to the presence of the lateral soft tissues which prevent lateral dislocations. While anterior and posterior dislocations have been reported clinically, they are less common than medial dislocations [6].

1.4.2.2 *Cause of dislocation*

The correct bearing thickness appropriately tensions the knee such that it just tightens the ligaments in extension. If the bearing is too thin, this could lead to knee instability and increase the likelihood of mobile bearing dislocation, particularly in flexion when the LCL is lax. If the selected bearing is too thick, it can lead to overstuffing of the knee joint, which has been shown to compromise clinical outcomes and also increase the bearing dislocation rate [160].

Some studies reported what the patient was doing when they think they dislocated and what they suspected was the cause of dislocation. Some patients reported a traumatic incident which may have led to their dislocation (e.g. a fall [39], falling off a bus [41], falling down the stairs [146], stepping into a rabbit hole [146], or squatting [162]). Sometimes, this traumatic incident was associated with damage to the ACL and LCL [6, 7, 163]. One study reported a medial dislocation which occurred while standing up from a seated position [163].

Dislocation is believed to be a complication of flexion and is more common laterally probably because the lateral ligaments become lax in knee flexion [164], thereby allowing the lateral compartment to distract and therefore making bearing dislocation more likely [32]. Other factors associated with knee flexion may also contribute to dislocation. For example, *in vivo* fluoroscopic studies of the ODL have shown that there is normal roll back of the femur on the tibia with flexion and that the amount of movement is greater than with a flat tibial component [165]. This lateral roll back is associated with marked rotation of the tibial component relative to the femoral component and also mediolateral displacement. Finally, a recent study found that as the knee flexes, the mobile bearing also overhangs the tibial component posteriorly by up to 50% [158].

1.4.2.3 *Treatment of dislocations*

Dislocations are usually associated with immediate symptoms, although the nature of these symptoms was not described, except in one case study as a “sharp pain” in the knee in which the dislocation occurred and a feeling of the knee “giving away” [163]. However, some dislocations are asymptomatic [40, 122] and, if diagnosed, have occasionally spontaneously resolved [122]. Usually, symptomatic dislocations have been successfully treated by bearing exchange whereby the dislocated bearing is removed and a new bearing, of the same thickness as the original bearing or thicker, is inserted [147]. In addition, the bearing and knee are examined for evidence of impingement of bone or cement on the bearing and if this is found, it is addressed [6, 7, 19, 40, 114]. There are occasional reports of closed reduction of the bearing under general anaesthesia and anecdotal reports of reduction under local anaesthesia [39, 41, 112]. One study successfully reduced a dislocated bearing through arthroscopic manipulation, wherein considerable varus stress was applied to the knee using a leg-holder and the bearing was manipulated using a hook palpator [163]. For primary dislocation, particularly if there are other problems, or for recurrent dislocations, while it is possible to convert to a TKR [96], it is probably best to replace with a fixed bearing UKR device such as the FLO.

With a rapidly ageing population and a greater demand for treatments for younger and more active patients, there is considerable demand for implants with greater longevity and longer survivorship. If the complication of dislocation of the mobile bearing could be resolved, mobile bearing implants like the ODL implant could outperform fixed bearing designs, not only as a result of its capacity for more normal knee kinematics [166] but because bearing dislocation tends to occur early, within 12 months post-operatively [19], which may mask the long term benefit of reduced bearing wear.

Chapter 2 BEARING DISLOCATION IN THE ODL IMPLANT – META-ANALYSIS COMPARING THE OXFORD FLAT LATERAL VS. THE ODL IMPLANT

The results from this chapter have been published in a peer reviewed journal[2]

2.1 Introduction

At the centre in Oxford, changing the design of the tibial component from a flat articular surface to a domed articular surface [159] appears to have decreased the incidence of dislocation [7]. The extent by which this innovation has decreased dislocation more widely is unknown. Therefore, this chapter presents the results of a systematic review and meta-analysis conducted to compare the rate of mobile bearing dislocation of the Oxford flat lateral implant and the ODL implant. This study was registered with PROSPERO: registration number CRD42019139250.

2.2 Methods

The population, intervention, comparators, outcomes and studies (PICOS) framework to inform the review objectives were defined as per Table 2.1.

[2] I. Yang, T. Hamilton, S. Mellon, D. Murray (2021), A systematic review on the dislocation of the mobile bearing in the Domed Lateral Oxford Knee: flat versus domed, vol 28, pp.214-28, The Knee.

Table 2.1. PICOS outline to define the systematic review question.

Population	Adult patients (>18y/o) with lateral compartment knee osteoarthritis refractory to non-surgical treatment.
Intervention	Surgical treatment using either the flat or the ODL implant.
Comparators	N/A
Outcomes	Primary outcomes - Dislocation rate (see methods for calculation method) Secondary outcomes - Dislocation mode: anterior, posterior, medial or lateral - Cause(s) of the dislocations - How the dislocation was resolved - Time of dislocation (early <12months post-surgery, medium >12months <2 years, late >2 years - Implant survival with dislocation as the endpoint Sub-group stratification categories Country of study Patient mean age Mean follow-up duration Surgeon type: designer versus independent Surgeon case-load
Studies	Included: Randomised Control Trials (RCTs), cohort studies, case control studies, case series Excluded: paediatric cohort (<18y/o), reviews, animal studies, mechanical model studies, cadaveric studies, editorials/expert opinions, case reports.

2.2.1 Search strategy

1. Arthroplasty, Replacement, Knee/
2. (partial knee or unicompartmental or unicondylar).ti,ab.
3. OR/1-2
4. partial.ti,ab.
5. unicompartmental.ti,ab.
6. unicondylar.ti,ab
7. uni.ti,ab.
8. (UKA or UKR or UCA or UCR or PKA or PKR or PCA).ti,ab.
9. meniscal.ti,ab
10. bearing.ti,ab.
11. dislocat*.ti,ab.
12. revis*.ti,ab.
13. re-operation.ti,ab.
14. reoperation.ti,ab.
15. surviv*.ti,ab.
16. Oxford.ti,ab.
17. failure.ti,ab.
18. mobile.ti,ab.
19. OR/4-18
20. 3 AND 19
21. (robot* or total or TKA).m_titl.
22. 20 AND 21
23. (partial or unicompartmental or unicondylar or uni or UKA or UKR or UCA or UCR or PKA or PKR or PCA).m_titl.
24. 22 not 23
25. 20 not 24
26.

Three electronic databases, MEDLINE, EMBASE and ISI Web of Science, were searched to identify studies reporting on the outcomes of mobile bearing ODL UKR (Zimmer Biomet, Warsaw, IN). The search strategy was implemented on the 15th March 2019 for MEDLINE and EMBASE and repeated on the 25th May 2020. For ISI Web of Science, the search strategy was implemented on the 22nd June 2020. The search strategy started with a relevant MESH term, followed by all synonyms for the concept of partial knee replacement and mobile bearing. Then, specific search terms regarding (1)

bearing dislocation, (2) failure, (3) bearing and (4) mobile were searched. Search strings were built in MEDLINE and subsequently modified for EMBASE. The search strategy was developed in conjunction with an experienced librarian, and the research team members who have experience in systematic reviews in this area of research. No date limit or language restrictions were imposed on the search results. The reference list of relevant systematic reviews captured by the search were also reviewed for eligible studies for inclusion. The reference list of the final list of included studies were also reviewed for eligible studies.

2.2.2 Selection criteria for studies

All studies containing patients treated with either the flat or the domed implants and their outcomes were included in the systematic review. These included: Randomised Control Trials (RCTs), cohort studies, case control studies, and case series.

All reviews were excluded. Any studies assessing radiographic imaging or conducted in animals, models (mechanical and computational), and *in vitro* were excluded from the review. Editorials/expert opinions were not included and case reports were also excluded because the assessment of dislocation rate was a primary aim. Studies on registry data were excluded due to insufficient data specific to lateral mobile bearing UKRs. Wherever possible, the speculative causes for dislocations, and other relevant findings were summarised and reported narratively under the discussion section.

2.2.3 Data collection

The findings were exported with full references to Endnote (version ≥X8, Clarivate Analytics) and duplicate entries were removed. Following this, titles and abstracts were independently screened and discrepancies were resolved by Irene Yang and Thomas W. Hamilton. All authors were then sent the final list of included studies for agreement. Data extraction of the included studies was conducted by Irene Yang and Thomas W. Hamilton independently with disputes resolved by Stephen J. Mellon or David W. Murray.

For each study, where possible, the following information was extracted: first author, publication date, country of study, study type, level of evidence, total number of knees operated, total number of patients, number of male patients, number of female patients, mean age (years), age range of patients, study start date, study end date, number of surgeons, surgeon type (either designer or independent), surgical technique used, mean follow up duration (months), follow up range (months), number of bearing dislocations reported, mode of dislocation (anterior, posterior, medial, lateral), cause(s) of the dislocations, how the dislocation was resolved, time of dislocation (months), implant survival with dislocation as the endpoint.

2.2.3.1 *Number of patients followed up*

With regards to the number of patients followed up, this was calculated by taking the total number of patients and subtracting the number of patients lost to follow up and those who had refused to participate. It was decided to include the patients who had died into the analysis as most clinical records would have information regarding patient knee revision status.

Where data was missing or incomplete, attempts were made to contact study authors to obtain this information where possible. All statistical analyses were conducted using Stata, version 13 (StataCorp, College Station, TX) with $p < 0.05$ considered statistically significant.

2.2.4 Strategy for data synthesis

2.2.4.1 *Statistical analysis*

As revision rates follow a binomial distribution, a meta-analysis of proportions was performed by Thomas W. Hamilton (*metaprop* command in Stata). Using the *metaprop* command ensured that studies with 0% or 100% proportions were included in the meta-analysis. The summary dislocation incidence or annual revisions were pooled using a random-effects model to minimise the effect of between-study heterogeneity. The I^2 statistic assessed the percentage of total variation due to inter-study heterogeneity.

The primary comparison between flat and domed tibial components was dislocation, which was assessed either as a rate (number of dislocations/number of cases, expressed as %) or as an annual rate (%pa) calculated by the Patient Time Incidence Rate (PTIR) method [12]. The PTIR analysis is based on the assumption that the event rate is constant over time. Previous studies have suggested that this is not the case for dislocation, as dislocations tend to occur in the first year. If this study has the same finding, then dislocation would be best assessed by rate rather than annual rate (PTIR). PTIR determines the number of events per 100 component years and it is calculated by dividing the number of events by the total number of cases multiplied by the average follow up and is expressed as a percentage.

The PTIR method was used to assess revisions as these tend to be relatively constant over time. The annual revision rate was calculated for two different endpoints: all cause revision and revisions excluding those occurring as a result of dislocation; with data grouped by flat or domed tibial component design. PTIR was also used for the subgroup analyses such as designer versus non-designer and gender (male versus female). Survival rates were estimated by multiplying the annual failure rate by the number of years for the survival estimate and then subtracting this from 100.

2.2.4.2 *Confidence interval*

For both rates and annual rates, the 95% Confidence Intervals (CIs) were calculated using the Clopper-Pearson or exact binomial method for a binomial proportion. This is computed by inverting the equal-tailed test based on the binomial distribution. This calculation also incorporates the Freeman-Tukey double arcsine transformation of proportions, which enables the within-study variability to be modelled using the binomial method, and functions to stabilise the variance. This was necessary since revisions due to bearing dislocation were expected to be low.

2.2.5 Risk of bias assessment

Using a modified method of assessing bias previously reported by Mohammad et al. [147], Campi et al. [167] and de Vos-Kerkhof et al. [168], the level of bias in each of the included studies was assessed. The modified method uses the Methodological Index for evaluation of Non-Randomized

Studies (MINORS) and an additional scoring system based on the presence of primary outcome reporting (A = clearly reported, B = non-reported/unclear) and the number of cases in the studies (A > 100, B = 51–99, C = < 50) [167, 169]. Studies with a MINORS score over 80% were considered at low risk of bias and those below 70% at high risk (except those with 3 or more "A's" in outcome reporting and study size [167]).

2.3 Results

Table 2.2. Database search results.

Database searched	Date searched	Number of outcomes
Medline (Ovid MEDLINE® Epub Ahead of Print, In-Process & Other Non-Indexed Citations, Ovid MEDLINE® Daily and Ovid MEDLINE®) 1946 to present	15th March 2019	4345
Embase 1974 to present	15th March 2019	3060
Manually added	15th March 2019	10
Medline (Ovid MEDLINE® Epub Ahead of Print, In-Process & Other Non-Indexed Citations, Ovid MEDLINE® Daily and Ovid MEDLINE®) 1946 to present	25th May 2020	4927
Embase 1974 to present	25th May 2020	3421
ISI Web of Science	22nd June 2020	4941

Searches identified a total of 7405 articles, with an additional 10 manually identified studies. After duplicates were removed, 5050 articles remained. After screening, 21 full texts articles were retrieved and assessed. From the assessment, 12 texts were excluded (see Table 2.3), leaving nine studies which met the inclusion criteria (Figure 2.1). The nine included studies consisted of three flat tibial component design series, and six domed tibial component design series. Of the three flat series studies, all were retrospective cohort studies. Of the six domed series studies, five studies were retrospective cohort studies and one prospective cohort study.

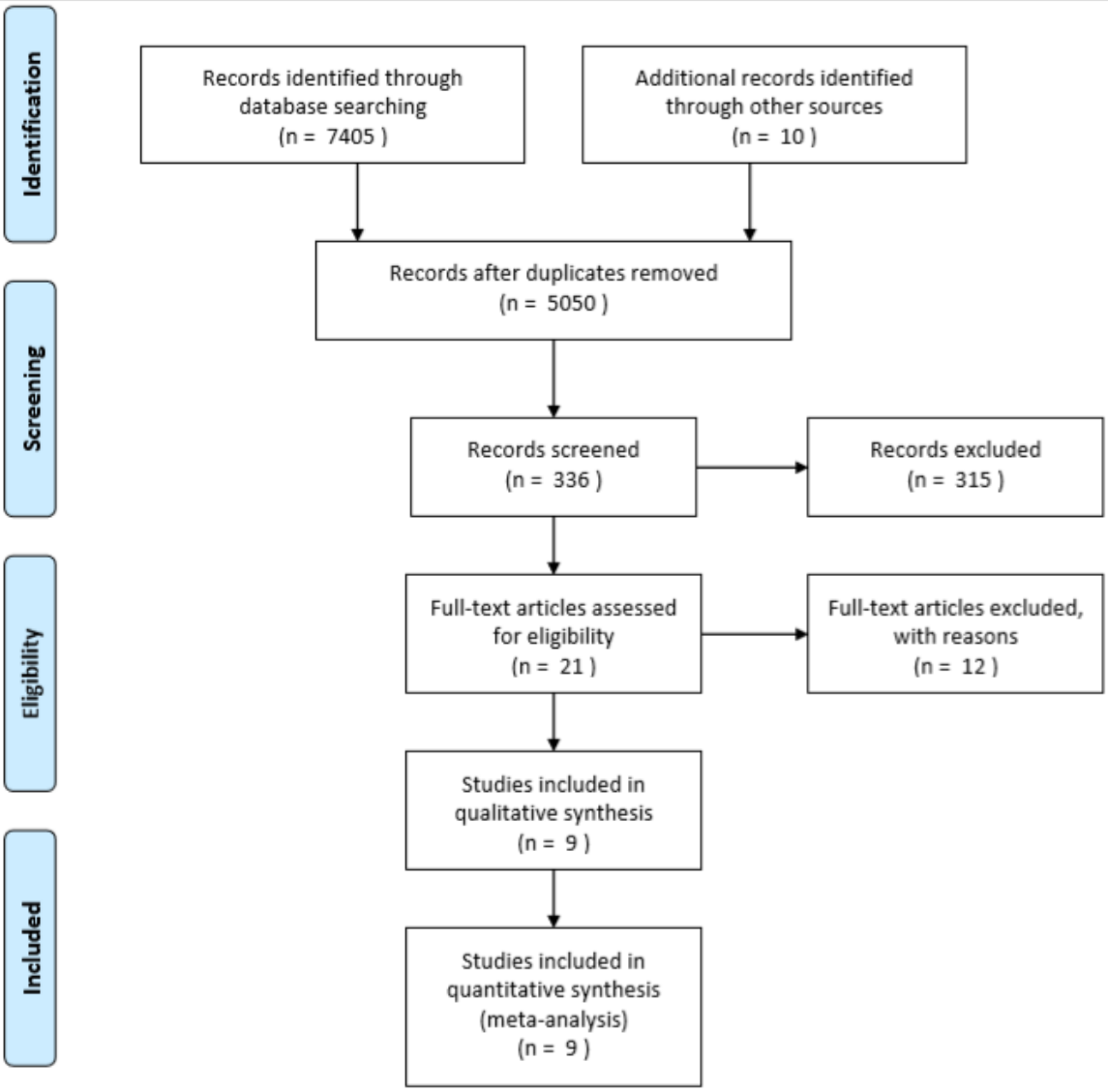


Figure 2.1. PRISMA flow diagram for study inclusion.

Table 2.3. Excluded studies with reasoning after reading full manuscript.

Author	Publication date	Country of study	Exclusion reason
Flat tibial component			
Goodfellow [90]	1986	UK	Cannot separate out data of medial versus lateral dislocations.
Goodfellow [170]	1987	UK	Not clear what information is lateral in this paper. Additionally, uses same cohort as Goodfellow (1988) paper.
Goodfellow [122]	1988	UK	Bicompartmental series.
Hangody [171]	1992	Hungary	Information is not split out for the lateral Oxford knee and it is a registry based study which was in the exclusion criteria.
Lewold [141]	1995	Sweden	Registry based study.
Domed tibial component			
Pandit	2010	UK	The first series is a repeat of Gunther (1996) without updates, and the third series is the same cohort as in Kennedy (2019).
Pandit [172]	2012	UK	Repeat of the Gunther (1996) and Pandit (2010) paper without updates.
Streit [96]	2012	Germany	The same cohort is reported in the Walker (2018) paper with updates.
Altuntas [155]	2013	UK	The same cohort reported in the Newman (2017) paper with updates.
Weston-Simons [146]	2014	UK	The same cohort reported in the Kennedy (2019) paper with updates.
Walker [113]	2014	Germany	No dislocation information.
Masud [114]	2018	UK	Insufficient data regarding mean follow up period.

Reasons for excluding studies are listed in Table 2.3. All studies were included in the quantitative analysis for the primary outcome of mobile bearing dislocation.

Table 2.4. Risk of bias assessment for included studies.

Author	Year	Country	Clearly stated aim	Inclusion of consecutive patients	Prospective data collection	Endpoints appropriate to study aim	Unbiased assessment of study endpoint	Follow-up period appropriate to study aim	<5% lost to follow up	Prospective calculation of study size	Adequate control groups	Contemporary groups	Baseline equivalence of groups	Adequate statistical analysis	MINORS score	Sample size	1° outcome reporting	Risk of bias
Flat tibial component																		
Kehr [173]	1995	France	1	2	1	0	0	2	2	1	N/A	N/A	N/A	N/A	9	C	A	H
Gunther [19]	1996	UK	0	2	1	0	1	2	2	0	N/A	N/A	N/A	N/A	8	B	B	H
Pandit [7]	2010	UK	1	2	1	1	1	2	2	0	N/A	N/A	N/A	N/A	10	B	B	H
Domed tibial component																		
Schelfaut [8]	2013	Belgium	2	2	2	1	2	2	2	2	N/A	N/A	N/A	N/A	15	C	A	L
Marson [112]	2014	UK	1	2	1	1	2	2	2	0	N/A	N/A	N/A	N/A	11	C	B	H
Newman [39]	2017	UK	2	2	2	2	2	2	2	1	N/A	N/A	N/A	N/A	15	B	B	L
Fornell [41]	2018	Spain	2	2	1	2	2	2	2	0	N/A	N/A	N/A	N/A	13	C	B	L
Walker [40]	2018	Germany	1	2	1	1	2	2	2	0	N/A	N/A	N/A	N/A	11	A	B	H
Kennedy [6]	2020	UK	2	2	2	2	1	3	2	1	N/A	N/A	N/A	N/A	15	A	B	L

Using the modified method of assessing bias, previously reported by Campi et al. [167] and de Vos-Kerkhof et al. [168], the level of bias in each of the studies included in quantitative data synthesis revealed five series at high risk of bias and four series with low risk of bias (Table 2.4).

Table 2.5. Demographic information of studies included in quantitative analysis.

Author	Year	Country	Study type	Total no. of knees	Total no. of patients	% Male patients (no.)	% Female patients (no.)	Mean age	Age range
Flat tibial component									
Kehr [173]	1995	France	Retrospective cohort	5	3	None stated	None stated	None stated	None stated
Gunther [19]	1996	UK	Retrospective cohort	53	51	8 (4)	92 (47)	68	40 – 88
Pandit [7]	2010	UK	Retrospective cohort	65	64	44 (28)	56 (36)	53	39 – 83
Domed tibial component									
Schelfaut [8]	2013	Belgium	Prospective cohort	25	25	36 (9)	64 (16)	60	31 – 86
Marson [112]	2014	UK	Retrospective cohort	15	None stated	None stated	None stated	58	41 - 77
Newman [39]	2017	UK	Retrospective cohort	64	58	29 (17)	71 (41)	71	44 - 92
Fornell [41]	2018	Spain	Retrospective cohort	41	41	29 (10)	71 (31)	63	38 - 81
Walker [40]	2018	Germany	Retrospective cohort	344	327	24 (90)	76 (237)	65	36 - 88
		Heidelberg*		120	114	28 (32)	62 (82)	None stated	None stated
		Bad Rappenau*		32	31	35 (11)	65 (20)	None stated	None stated
		Stuttgart*		192	182	26 (47)	74 (135)	None stated	None stated
Kennedy [6]	2020	UK	Retrospective cohort	325	308	34 (104)	66 (204)	65	39 - 90
TOTAL				937	877	30 (262)	70 (612)	63 (64**)	N/A

**Sub study breakdown by location within Germany.*

***Mean weighted age. Excluded Kehr et al. study due to insufficient information.*

From the data in Table 2.5, more women (70%) suffer from lateral compartment disease than men (30%).

Table 2.6. Dislocation with patient specific information available.

Dislocations with dislocator age info. Available (%*)	21 (35)
Average age of dislocators in years (range)	61 (39 – 84)
Dislocations with sex info. available (%*)	22 (37)
Number male dislocators (%)**	9 (41)
Number female dislocators (%)**	13 (59)
Mean time of dislocation (n=28) (range) (months)	16 (0.6 – 104.4)

*Percentages have been calculated with respect to the total number of dislocations (n=60) across all studies.

**Percentage out of the total who have reported

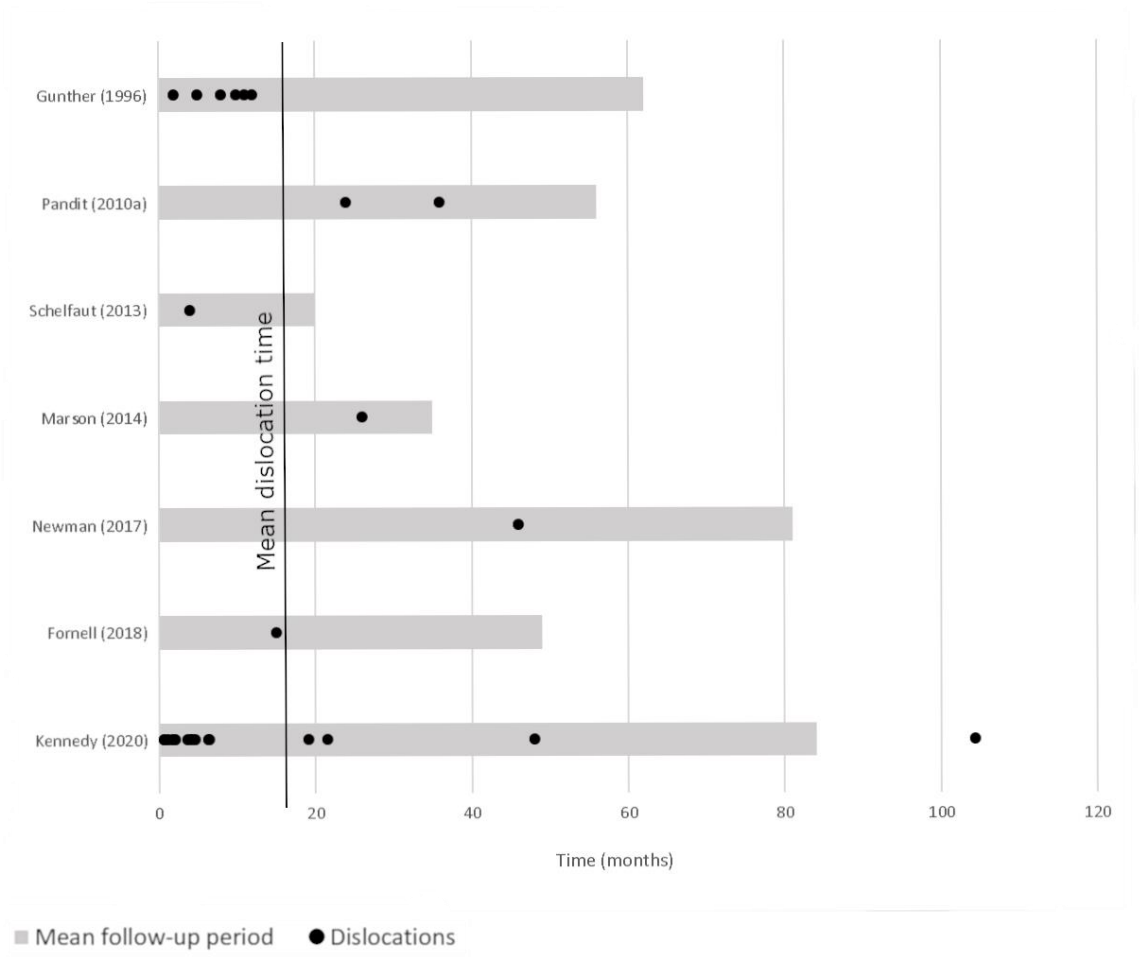


Figure 2.2. Time of first dislocation(s) with respect to mean follow up duration in months. The mean time of first dislocation (16 months) is marked and labelled.

In Table 2.6, using the 21 dislocations with age reported (35% of total number of dislocations), the average age of patient with a dislocation was 61 years old (range 39 – 84). Similarly, using the 22 dislocations with gender reported (37% of total number of dislocations), more women (59%) dislocate than men (41%).

Figure 2.2 shows the studies that reported information about time of dislocation (8 studies reporting on 28 dislocations). From these studies, the average time of first dislocation is 16 months post-op (range 0.6 – 104.4 months). Given an average follow up of 50 months (range 20 – 84), dislocations seem to occur relatively early.

Table 2.7. Mobile bearing dislocation data for all studies reporting no. of dislocations.

Author (Year)	Total no. of knees	Mean follow up duration (months)	No. bearings dislocated	Dislocation risk (%)	No. of asymptomatic bearing subluxations	Dislocation mode (% of the total)
Flat tibial component						
Kehr (1995) [173]	5	39	4	80	None Stated	None stated
Gunther (1996) [19]	53	62	6	11	None Stated	None stated
Pandit (2010) [7]	65	56	3	5	None Stated	67% medial dislocations into the intercondylar notch.*
Domed tibial component						
Schelfaut (2013) [8]	25	20	1	4	None Stated	100% Medial dislocation into the intercondylar notch.
Marson (2014) [112]	15	35	1	7	None Stated	None Stated
Newman (2017) [39]	64	81	1	2	None Stated	None Stated
Fornell (2018) [41]	41	49	1	2	None Stated	100% Medial dislocation
Walker (2018) [40]	344	37	19	6	1	None Stated
Kennedy (2020) [6]	325	84	14	4	None Stated	71% medial dislocations 14% unknown 14% anterior dislocations

**Remaining dislocation directions were not mentioned.*

From the clinical reports included in this meta-analysis, in the four papers where the mode of dislocation was reported (Table 2.7), all had indicated that at least one of the dislocations was medial into the intercondylar notch with the bearing lodged above the wall of the tibial component. Of the flat series, it was only the Pandit et al. [7] publication, which stated the mode of dislocation and this was reported as medially. Of the nine studies comprising the domed series, three studies [6, 8, 41] had reported the mode of dislocation as medial dislocations, aside from two anterior dislocations reported by Kennedy et al. [6].

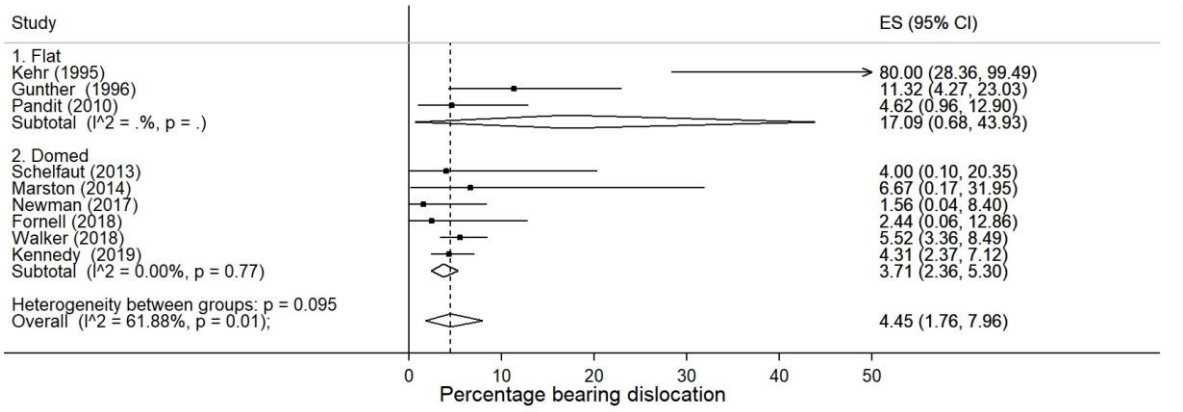


Figure 2.3. Percentage bearing dislocation assessment of flat versus domed tibial component design.

Forest plots were generated showing the individual effect sizes and the overall effect size (ES). The overall incidence of bearing dislocation in both flat and domed tibial designs was 4.5% (Figure 2.3). Changing the design of the tibial component from flat to domed decreased the dislocation rate from 17% (95%CI: 0.68 – 43.93) in the flat to 3.7% (95%CI: 2.36 – 5.30) in the domed series. The large confidence interval seen in the flat series is largely attributed to the Kehr et al. (1995) study [173]. Removing the Kehr et al. study from the analysis reduced the dislocation rate for the flat tibial tray design to 7.3% (95%CI: 3.08 – 12.91), but this was still higher than that for the domed component (Figure 2.4).

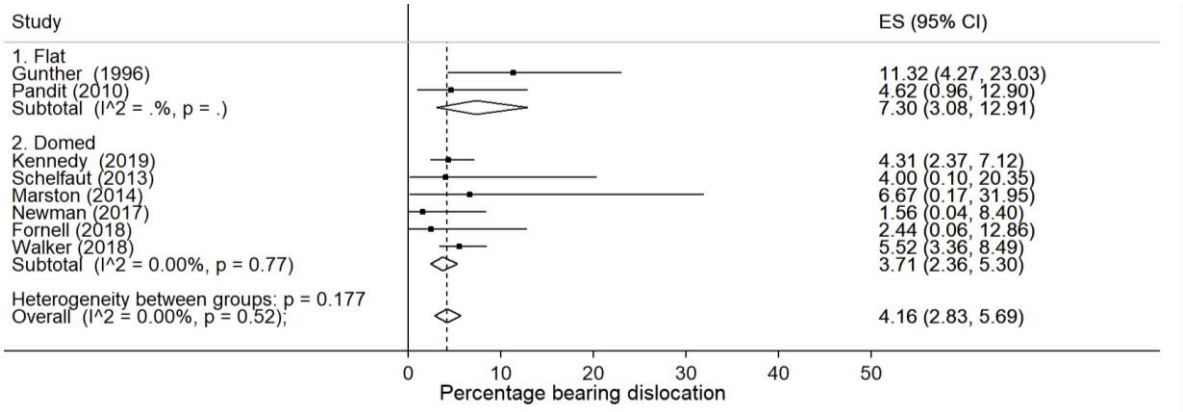


Figure 2.4. Percentage bearing dislocation assessment of flat versus domed tibial component design – excluding Kehr et al. study.

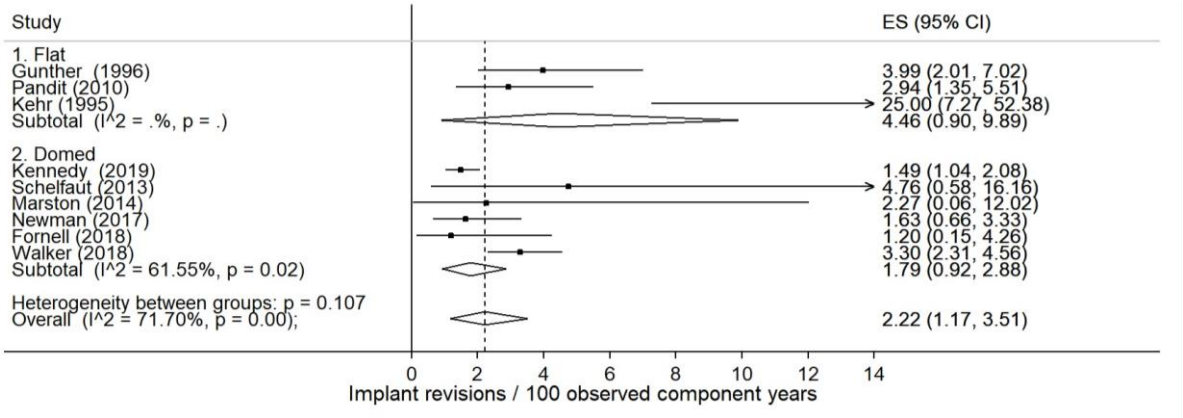


Figure 2.5. PTIR assessment for all cause revision: Flat versus domed tibial component.

The PTIR for all cause revision for the flat tibial tray was 4.5%pa, equivalent to a five-year revision rate of 22% (Figure 2.5). For the domed tibial tray, it was 1.8%pa, equivalent to a five-year revision rate of 9.0%.

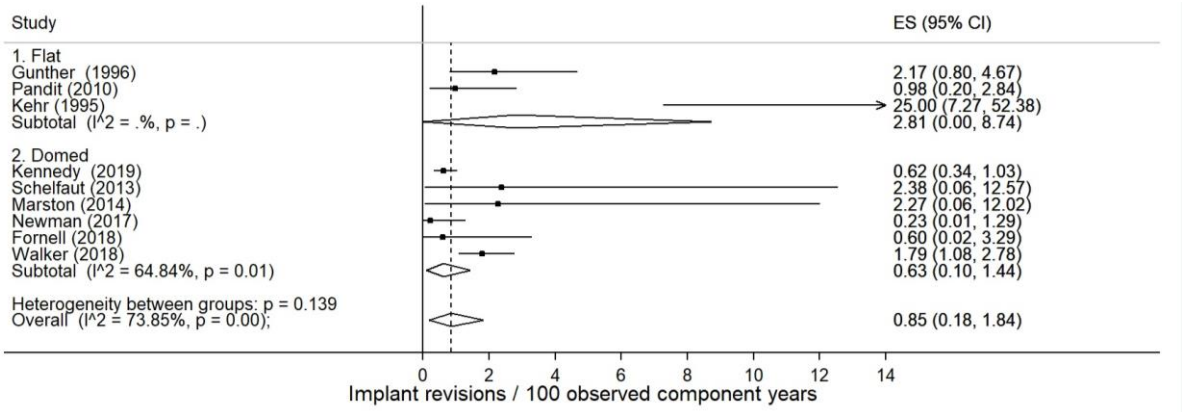


Figure 2.6. PTIR assessment of bearing dislocation: flat versus domed tibial component design.

The PTIR for bearing dislocation for the flat tibial tray was 2.8%pa, equivalent to a five-year dislocation rate of 14% (Figure 2.6). For the domed tibial tray, the PTIR for dislocation was 0.6%pa, equivalent to a five-year dislocation rate of 3%.

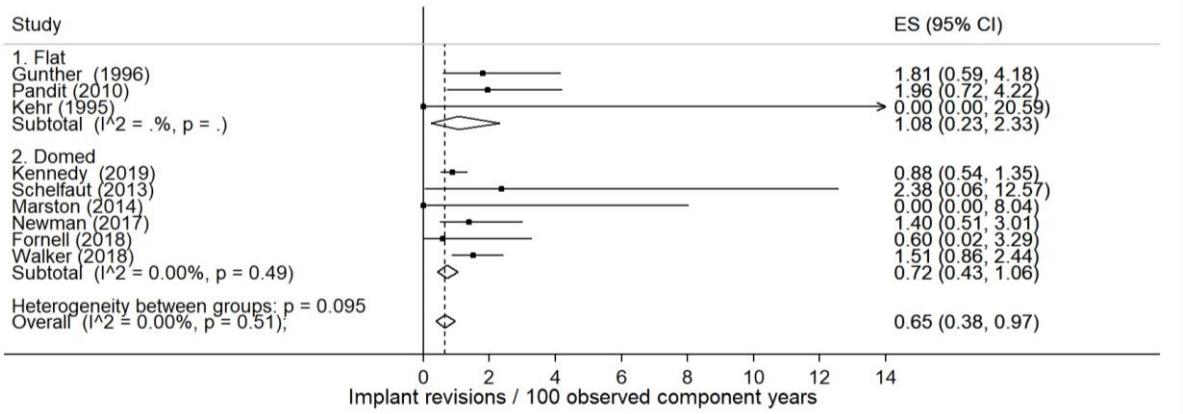


Figure 2.7. PTIR assessment for all cause revision except bearing dislocation: Flat versus domed tibial component.

The PTIR for revision excluding dislocation for the flat tibial tray was 1.1%pa, equivalent to a five-year revision rate of 5.5% (Figure 2.7). For the domed tibial tray it was 0.7%pa, equivalent to a five-year revision rate of 3.6%.

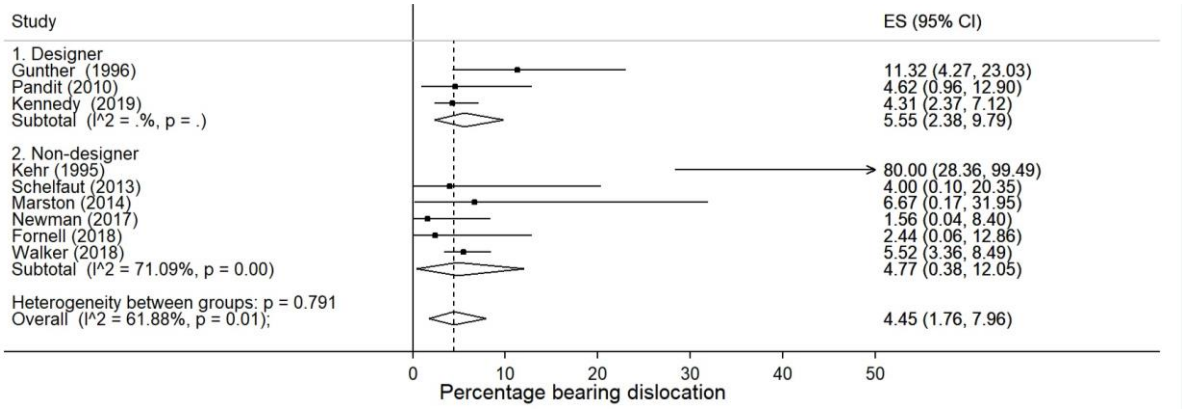


Figure 2.8. Percentage bearing dislocation: designer versus non-designer.

Subgroup analysis by designer versus non-designer showed no difference (p=0.791) (Figure 2.8).

2.4 Discussion

This study identified that changing the design of the tibial component from a flat tibial tray to a domed tibial tray reduced the dislocation rate from 17% to 3.7%. There was also an associated decrease in revision rate, and the proportion of revisions that were due to dislocation decreased from

two thirds with the flat tibial component to one third with the domed tibial component. Further, the study found no difference in dislocation rate between designer and non-designer surgeons and found information to better characterise the problem of mobile bearing dislocation. Whilst the reduction in dislocation rate has been attributed to changing the tibial component from a flat to a domed tibial surface design, concurrent changes over time to the patient selection, instrumentation, surgeon experience and surgical technique were not accounted for.

2.4.1 Subgroups or subset analyses: deviations from PROSPERO

Subgroup analyses were conducted to compare the effect estimate between studies to evaluate the effect of sex (male versus female), geographic location of studies, and surgeon type (designer versus non-designer) on the rate of mobile bearing dislocation. Unfortunately, there was insufficient data to adequately assess the effect of geographical region and therefore, ethnicity and the impact of genetics. There was not a good spread of information regarding the geographical region, and most large studies published originated out of the designer surgeons based in the UK. Similarly, there was insufficient data to adequately assess the effect of surgeon caseload on the rate of mobile bearing dislocation.

2.4.2 Dislocation and revision rates

Whether dislocation was assessed by the overall rate or the annual rate (PTIR) the dislocation rate was about one fifth with the domed compared to the flat tibia. Dislocations tended to occur early (relative to the total follow up time) with the average time to dislocation being 16 months post-operatively. Therefore, the best way to assess dislocation is with the overall rate rather than the annual rate. The overall dislocation rate for the domed tibia was 3.7%.

As the revision rate tends to be relatively constant, revision is best assessed by the annual revision rate (PTIR). The change from flat to domed tibia decreased the annual revision rate by about two thirds and decreased the annual revision rate, excluding revisions for dislocation, by about one third. This suggests that the domed tibia has other advantages beyond dislocation compared to the flat, which probably relate to its more anatomical shape and better kinematics. Although it was not

possible to compare patient functional outcome in this study the evidence from the literature suggests that the change from flat to domed did not compromise patient outcome and the domed tibia has achieved consistently good results when assessed with the Oxford Knee Score (OKS) [7]. Further, there is evidence to suggest the functional results of the domed were better than those of the flat, with an improved range of movement and more normal kinematics [165].

The annual revision rate, excluding dislocation, for the dome tibia was 0.7% pa, which equates to a ten year survival of about 93%. With an additional 3.7% revision rate due to dislocation the ten year survival would be about 90%, which is acceptable and not dissimilar to that of fixed bearing UKR [151]. However, if the dislocation rate could be decreased further it would be beneficial. The ultimate goal would be to reduce the rate of mobile bearing dislocation in the lateral compartment to match that of the medial compartment (0.7%) [174].

2.4.3 Additional information on dislocations

Whilst there was insufficient data to ascertain the definitive cause of dislocation, all dislocated knees were reportedly in a position such that the lateral compartment of the knee was unloaded and in flexion (subject to trauma), and therefore, could be distracted. Although women were more likely to have lateral UKR than men (70% versus 30%) (Table 2.5), the risk of dislocation was not related to sex. There is insufficient evidence to determine the precise rate of recurrent dislocation, but the evidence that exists suggests that it is about one in three [6, 19, 39, 40, 122].

2.4.4 Study limitations

There are a number of limitations with this study. Firstly, isolated lateral unicompartmental disease is relatively rare, accounting for 10% of all arthritic knees [4], therefore, few lateral UKR are implanted and few publications were available for inclusion in the systematic review. As a result there were only 937 knees in this review, so it was inadequately powered for formal statistical comparisons between groups and it was required to deviate from the proposed PROSPERO analysis (see 2.4.1 Subgroups or subset analyses: deviations from PROSPERO). Furthermore, publications are likely to be from high volume centres, which may limit the generalisability of the results.

However, there was no evidence that the designer surgeons, who published many of the papers, had different results to the non-designer surgeons.

This study assessed dislocation both in terms of the absolute rate and the annual rate. It also assessed revision based on the annual rate. The annual rate was calculated using the PTIR method in accordance with routine practice of the National Joint Registry (NJR) [5]. Due to the limited data it was not possible to do a formal survival analysis or to assess functional outcomes. Due to the low number of studies available, confidence in the dislocation rate estimate reported is limited due to the low volume of evidence available and the true dislocation rate may be different from the estimate provided here. Further, it was decided not to eliminate studies based on the level of bias nor to calculate the overall bias across studies due to the limited benefit this would bring. Instead, in addition to reporting the dislocation results for all studies, the dislocation results after selectively eliminating the Kehr et al. study [173] are also presented because it was an outlier with four dislocations out of five operated knees, which is far higher than any of the other included studies. The high dislocation rate is probably the result of inappropriate surgical technique and including this study in the meta-analysis may have disproportionately swayed the results in favour of the domed tibial component series. However, despite removing the Kehr et al. study, the dislocation rate of the dome was still half that reported for the flat.

There was limited data for outcomes of interest: country of study, surgeon case-load, male versus female dislocation. Therefore, it was decided to summarise this information reported, but not to use it for subgroup analysis.

Wherever possible, caution was exercised in data interpretation and when reporting conclusions.

2.5 Conclusions

In this study, it has been shown that changing the design of the mobile bearing lateral UKR tibial component from flat to domed has reduced the bearing dislocation rate from 17% to 3.7%. With the option to change from mobile to fixed bearing intra-operatively if there is marked bearing instability, the dislocation rate should drop further. This study also showed that the change to the domed tibia

reduced the revision rate excluding dislocation by about one third. With the domed tibia the annual revision rate, excluding dislocation, was 0.7% pa, which equates to a ten year survival of about 93%.

2.6 Acknowledgements

Hasan R. Mohammad from the Oxford Orthopaedic Engineering Centre (OOEC) and Catherine Hartley of the Nuffield Orthopaedic Library provided feedback and guidance for the design of the search strategy. Jules Morel assisted with translation of one included study.

Chapter 3 THE EFFECT OF COMPONENT POSITION ON THE RISK OF BEARING DISLOCATION: A MECHANICAL STUDY WITH A CUSTOM-DESIGNED RIG

The results from this chapter have been submitted to a peer reviewed journal for consideration[3].

3.1 Introduction

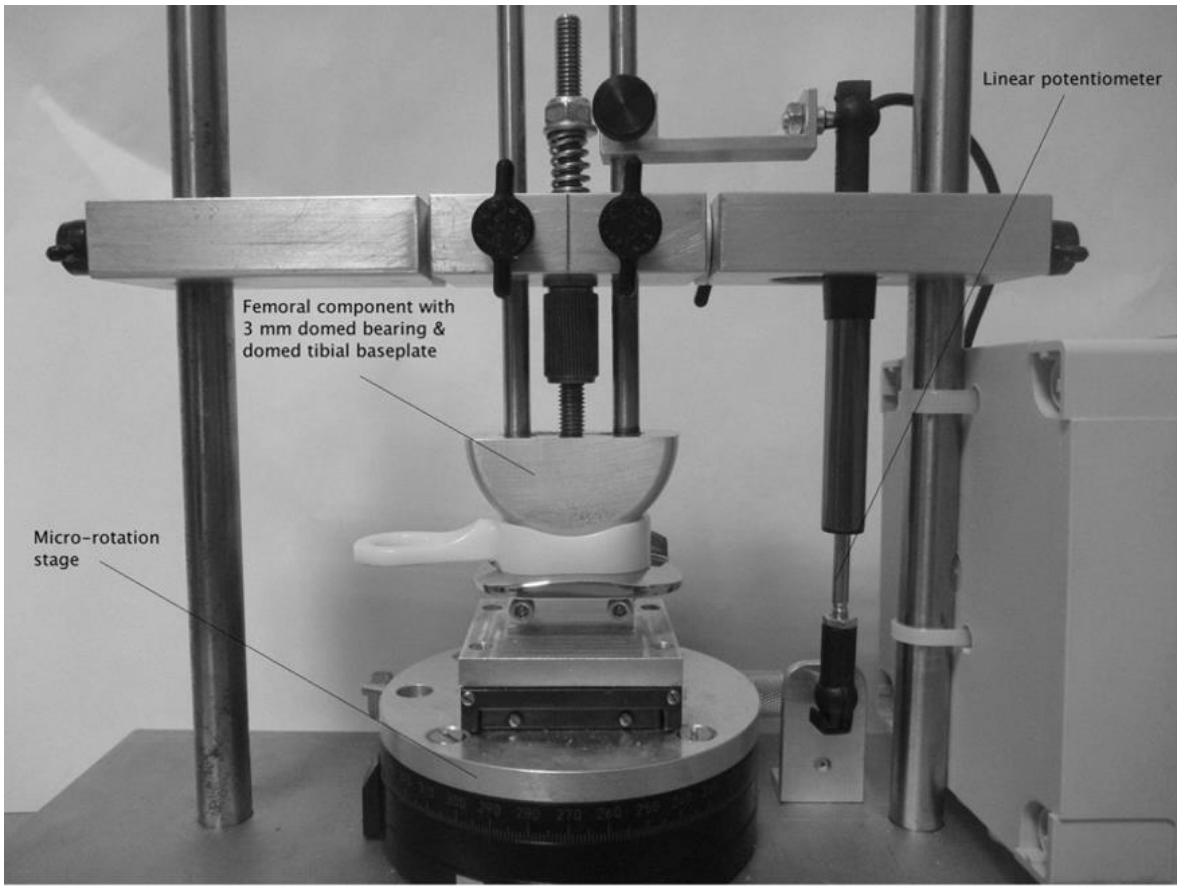


Figure 3.1. Mechanical rig model the Oxford Domed Lateral implant femoral component fixed in neutral, a domed tibial component and 3 mm mobile bearing [10].

In 2014, Weston-Simons et al. described a custom built mechanical rig (Figure 3.1) to assess dislocation of the mobile bearing [10]. The rig featured a size medium femoral component, a size C

[3] I. Yang, G. Agustoni, D. Murray, S. Mellon (2023), Mechanisms of mobile bearing dislocation in lateral unicompartmental knee replacement. Under review at the time of submission.

tibial component and in between these two components a mobile bearing modified with a handle. In the rig, the femoral component is capable of vertical distraction and abduction/adduction, and the tibial component is capable of mediolateral (ML) translation, anterior/posterior (AP) translation and internal/external rotation. Vertical distraction moved the femoral component away from the tibial component using an aluminium crossbar for crude motion and a spring loaded stage micrometer (precision 0.01 mm, Mitutoyo Ltd., Japan) for finer motion control. The aluminium crossbar was attached to a linear potentiometer (Model PS-C95P-25, Strainsense, Potterspury, UK), which when used in conjunction with a standard voltmeter and DC wave generator, could be used to measure the linear displacement of the femoral component away from the tibial component. The tibial component is fixed to a base stage capable of mediolateral (ML) translation (precision 0.01 mm, Mitutoyo Ltd., Japan), which is mounted above a rotation stage (model MR85-AR, GMT shop, GMT Europe GmbH, UK) to allow for tibial component rotation. Using the rig, the femoral component can be progressively distracted vertically away from the tibial component until the bearing can dislocate. The amount of distraction required for a dislocation was termed the "Distraction to Dislocation" (DD). The larger the DD, the lower the risk of dislocation.

In Chapter 2 it was found that most clinical dislocations occur medially and with the knee in a flexed position. As the knee flexes, the femoral component rolls back relative to the tibial component [2] causing the bearing to overhang the back of the tibial component [21] (Figure 3.2).

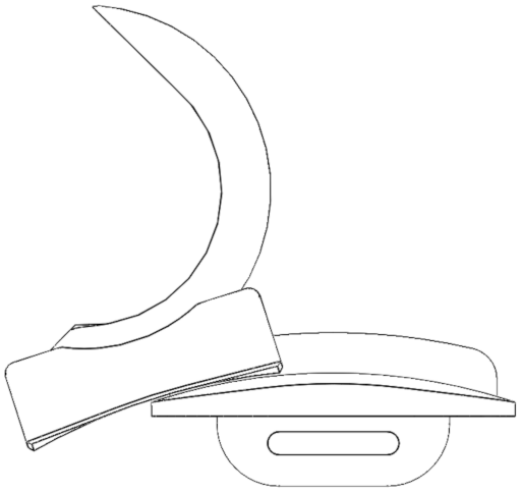


Figure 3.2. Schematic depicting how knee flexion can result in up to 50% of the bearing overhanging the tibial component posteriorly (relative to the right knee).

A recent study by Martin et al. [158] found that the amount of bearing overhanging the tibial component posteriorly can be up to 50% of the bearing length (Figure 3.3).

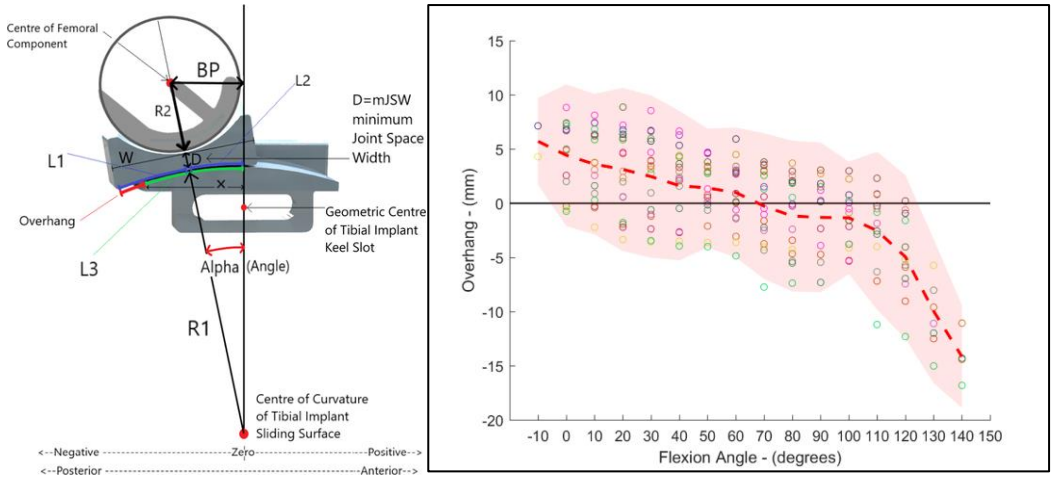


Figure 3.3. Bearing overhang (mm) (relative to the tibial component) versus flexion angle (°) for the ODL implant [158].

In addition to bearing overhang, the tibial component internally rotates relative to the femoral component with increasing knee flexion. Further, as there is more movement in the lateral as compared to the medial compartment of the knee [27], the bearing in the lateral knee tends to rotate around the medial compartment on a curved track so the bearing also moves mediolaterally [10].

3.2 Aims

The aim of this chapter was to use the findings from the Martin et al. study [158], to modify the mechanical rig described by Weston-Simons et al. [10] so that the rig can be used to mimic the relative position of the femoral and tibial components when the knee flexes (Part A). A subsequent aim of this chapter was to use the modified rig from Part A to study the effect of changing the tibial component rotation, posterior bearing overhang and mediolateral translation on the risk of bearing dislocation (Part B). It was hypothesised that while distraction is required for dislocation, the exact amount of distraction would vary depending on the relative positions of the implant components.

3.3 Chapter 3 Part A: Modifying the mechanical rig to mimic knee flexion

Before any modifications were carried out on the mechanical rig described by Weston-Simons et al. [10], the functionality of the rig was first assessed. The rig was first calibrated following the methods described by Weston-Simons et al. [10]. In brief, a standard wave generator was connected to channel 1 terminal of the mechanical rig. The ISO-TECH arbitrary function generator 5MHz USB (AFG-21005, RS pro) was set to a sine wave, DC output with an amplitude of 1 Vpp (peak-to-peak voltage) and a frequency of 10Hz. The output channel was connected to a digital multimeter (Silverline®). Using aluminium plates of known thicknesses (Figure 3.4), calibration measured the correlation between voltage readings (V) and vertical distraction distance (mm) (Table 3.1).

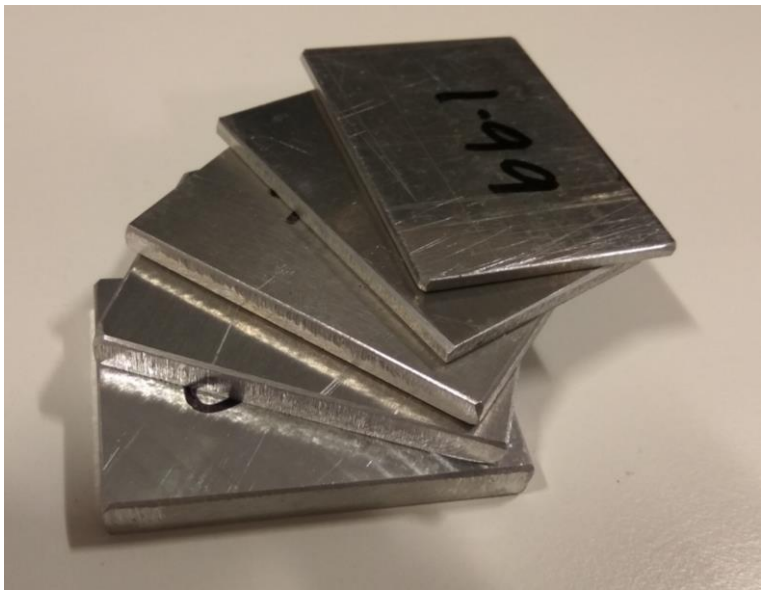


Figure 3.4. Metal calibration plates of thicknesses 2 mm, 3 mm, 4 mm, 5 mm and 6 mm.

Table 3.1. Voltage output readings for the known plate thicknesses.

Calibration block thickness (mm)	Actual measured thickness (mm)	Average voltage reading (V) (n=10 repeated measurements)
2	1.98	0.489
3	2.89	0.474
4	3.99	0.442
5	4.99	0.420
6	6.03	0.390

Using the readings from Table 3.1, a plot was generated and a linear line of best fit calculated (Figure 3.5).

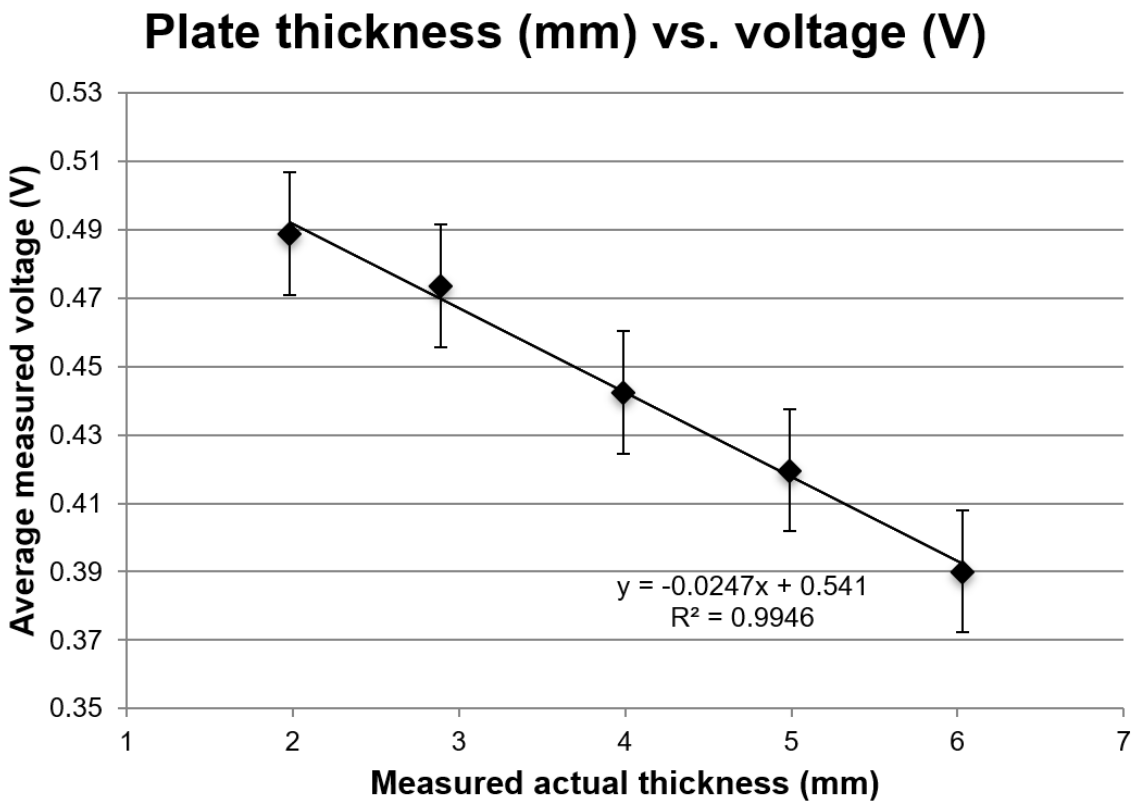
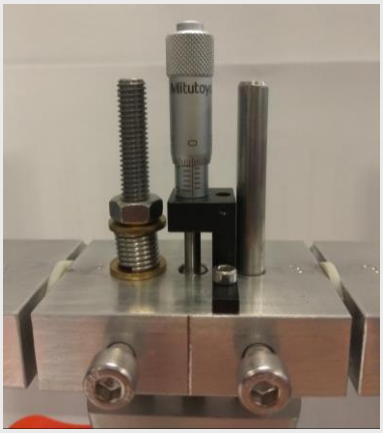
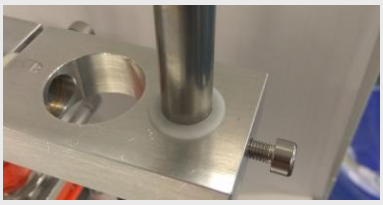
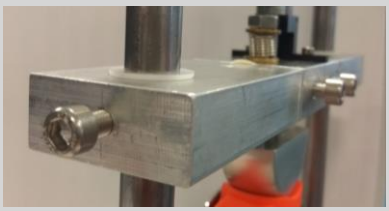


Figure 3.5. Scatter plot of plate thickness (mm) versus measured voltage (V).

Despite the high correlation coefficient of 0.9946 (Figure 3.5), it was found that the voltage readings between each 1 mm increment were numerically very similar which meant that the readings were not clearly discernible from one another. Furthermore, the voltage readings were unstable exhibiting constant fluctuations even if sufficient time was given between readings to allow them to stabilise. Finally, a little bit of play in the aluminium crossbar was noticed which could have affected the results. All of these factors lowered the confidence when considering the next step of optimizing the voltage readings for increments smaller than 1 mm. It was noted that fluctuations in the voltage readings could have resulted from the wave generator and its associated cables, the voltmeter, or the electronics linked to the potentiometer of the mechanical rig. However, the wave generator used was checked to be in date and stable and the voltmeter used for readings was brand new. The fluctuating voltage readings could have resulted from a faulty cable, so different cables were tested and all

connections checked, without success. Therefore, it was concluded that the fluctuating voltage readings probably resulted from faulty electronics linked to the linear potentiometer on the mechanical rig itself, which was likely given that the rig had been untouched for years. Instead of investigating the electronics, it was decided that the linear potentiometer would be replaced with a micrometer to simplify the system. The micrometer was accurate to 0.01 mm which ensured that accurate and stable readings for vertical displacement could be recorded. As the accuracy of a surgeon has been found to be no more than 1 mm [175], 0.01 mm accuracy for the tool was deemed sufficient for the purpose of the tool. Other minor changes were also carried out as detailed in Table 3.2 also to either aid in recording more reliable results, or to improve the use of the rig.

Table 3.2. Modifications made to the mechanical rig.

No.	Modification description	Image
1	Potentiometer removed.	N/A
2	A micrometer (Mitutoyo Ltd.) was attached to the top of the mechanical rig to enable accurate measurement of distraction distances.	
3	The central screw with spring was shifted from the centre to the left hand side. The spring itself was replaced with a new spring with lower stiffness. The spring feature was retained to ensure controlled movement of the femoral component.	
4	Plastic sleeves were added to the crossbar to better constrain motion of the crossbar to vertical distraction purely.	
5	Grub screws were removed and replaced with normal hex screws in order to enable tightening using an allen key.	

3.3.1 Aims

The aim of Part A of this chapter was to modify the mechanical rig to mimic the relative position of the femoral and tibial components when the knee flexes. The second aim was to verify that the changes made to the mechanical rig did not influence the rig performance by validating the results against those previously reported by Weston-Simons et al. [10].

3.3.2 Methods

3.3.2.1 *Modifying the rig to mimic knee flexion*

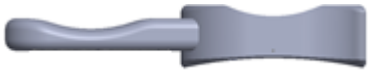
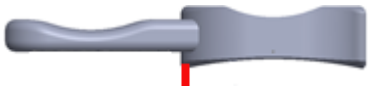
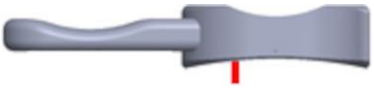
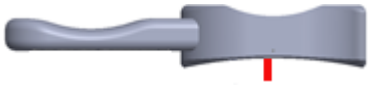
After, and in addition to, simplifying the design of the mechanical rig to enable the rig to record accurate and reliable DD readings, the rig was also modified to mimic the relative position of the femoral and tibial components when the knee flexes.

Based on the study by Martin et al. [158], the AP translation of the mechanical rig was modified to mimic the mean relative component positions at various flexion angles (Figure 3.3). Flexion angles of 0°, 90° and 130° were chosen to cover the range of physiological knee flexion. At 0° of knee flexion, the bearing was central on the tibial component (i.e. underhanging the posterior edge of the tibial component by 6.5 mm). At flexion angles of 90° and 130° the bearing was overhanging the tibial component edge posteriorly by 1 mm and 10 mm, respectively. In addition, the maximum bearing overhang of 50% was also mimicked, which corresponded to a flexion angle of 140°. For anterior-posterior (AP) translation the tibial component was tilted so that the vertical distraction was perpendicular to the tibial surface directly under the centre of the femoral component.

In order to model posterior bearing overhang at various flexion angles, it was required to modify the rig. As the femoral component was fixed, the tibial stage was modified so that the tibial component could rotate along a 75 mm spherical radius to match the tibial dome surface radius of curvature. Modifications were carried out using 3D CAD modelling software package, SOLIDWORKS 2020 (Dessault Systemes SOLIDWORKS Corp., Waltham, MA, USA).

As the modified bearing used in Weston-Simons et al.'s study [10] was no longer available, a 3D Computer Aided Design (CAD) model of the size 3 mobile bearing component was acquired from Zimmer Biomet (Swindon, UK). For testing, a bearing similar to the one described in Weston-Simons et al.'s study [10] was modelled. To setup for overhang testing, bearings with a tab positioned to correspond with 1 mm, 10 mm or 17 mm (50%) of bearing overhang were also modelled (Table 3.3). All bearings that were modelled featured a handle similar the one described in Weston-Simons et al.'s study [10].

Table 3.3. Modified mobile bearings designed. The bearing without any tabs is the bearing used for testing. The remaining bearings are used to setup overhang testing - red shows where the tabs were placed.

Amount of posterior mobile bearing overhang (mm)	Corresponding flexion angle (°)	Image
0	0	
1	90	
10	130	
17	140	

Using a 3D printer (Ultimaker 2 extended, Ultimaker B.V., The Netherlands), the modified bearings and the rotational platform were 3D printed in Polylactic Acid (PLA). Three copies of the modified bearing used for testing (with no tab) were printed to allow for worn and damaged parts to be replaced during testing. Using the 3D printed parts, the final rig was constructed with the help of James Fisk at the Institute of Biomedical Engineering (IBME) at the Department of Engineering Science at the University of Oxford.

3.3.2.2 *Validating the modified rig results: comparison with Weston-Simons et al.'s study*

To validate that changes made to the mechanical rig did not influence the rig performance, the dislocation testing from Weston-Simons et al. study [10] was repeated.

The starting position for testing was chosen with the bearing flush against the wall of the tibial component (Figure 3.6) and central on the tibia with 0° rotation and 0 mm ML translation. To set up for testing, the configuration was determined by setting the bearing AP position, tibial rotation angle and/or ML translation distance (the gap between the bearing and the tibial wall). For each configuration, the dislocation risk was assessed by recording the DD.

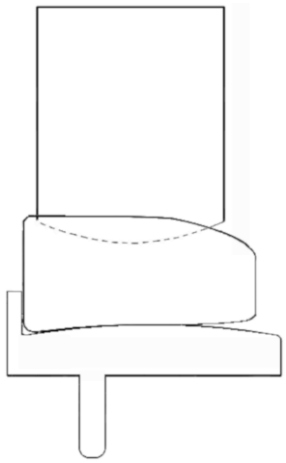


Figure 3.6. The starting position showing the bearing flush against the wall of the tibial component and with the femoral component resting on the superior surface of the bearing.

Between each testing configuration which varied the ML position, the vertical starting position was re-zeroed. The final testing parameters used to validate the results from Weston-Simons' study are shown in Table 3.4.

Table 3.4. Weston-Simons et al.'s validation testing parameters.

Testing parameter	**Testing range (increments)/values**
Bearing overhang values	Implant is central (no bearing overhang)
Rotational angles	0°
Mediolateral translation (increment)	0 – 8 mm (1 mm increments)
Vertical distraction (increment)	0.00 – 8.00 mm (0.25 mm increments)

Weston-Simons et al. [10] reported vertical distraction in the knee joint to an accuracy of 100 µm and Tokuhara et al.'s study reported the maximum vertical distraction in the lateral knee to an accuracy of 100 µm [32]. However, while knee distraction values less than 1 mm hold little meaning, it was decided that since Weston-Simons et al. [10] reported vertical distraction standard deviation values accurate to 10 µm, it was decided to conduct vertical distraction testing to an accuracy of 0.25 mm as a compromised accuracy. Since surgeon accuracy has been reported to an accuracy of 2.0 ± 1.3 mm [175], it was decided to conduct testing at 1 mm increments. All data were reported to two decimal places to allow comparison. To assess dislocation, the mobile bearing was manually manipulated using the attached handle. Dislocations were assessed in the three dislocation directions: medially over the tibial wall, anteriorly and laterally.

Medial dislocation

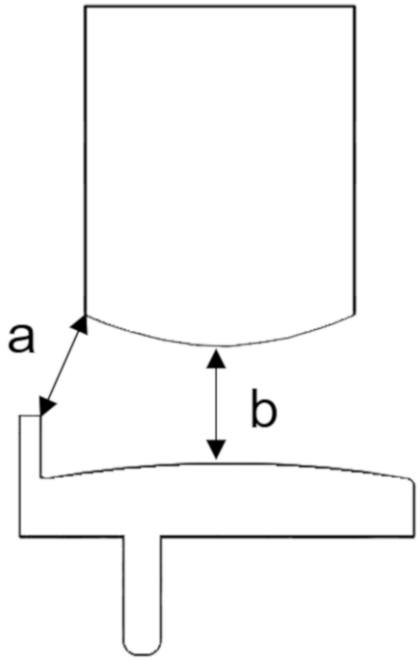


Figure 3.7. The minimum distance of the femoral component to tibial wall "a" and the minimum distance from the femoral component to the domed tibial surface "b".

Based on Weston-Simons et al.'s study [10], medial dislocations resulted in the bearing dislocating from between the metal femoral and tibial components. This occurred when the lateral side of the bearing cleared the minimum distance of the femoral component to tibial wall ("a" in Figure 3.7).

To dislocate the bearings medially, the bearing was allowed to dislocate using the manoeuvre specified by Weston-Simons et al. [10]: "*Firstly, the bearing was translated in an anterior direction and extended relative to the femoral component, maintaining full congruence with the femoral component. Secondly, it was internally rotated into valgus until the antero-medial corner was on top of the wall. Finally it was externally rotated and translated medially until the medial edge of the bearing was on top of the wall".*

Anterior dislocation and lateral dislocation

Anterior dislocation and lateral dislocation occurred when the bearing was manipulated until the bearing dislocated either anteriorly or laterally. Dislocations results in the bearings coming completely out from between the femoral and tibial components in the respective directions.

3.3.3 Results

3.3.3.1 *Modifying the rig to mimic knee flexion*

This modified stage design depicted in Figure 3.8 enabled the tibial component to be mounted on the top base allowing it to rotate along a 75 mm spherical radius to match the tibial dome surface radius of curvature.

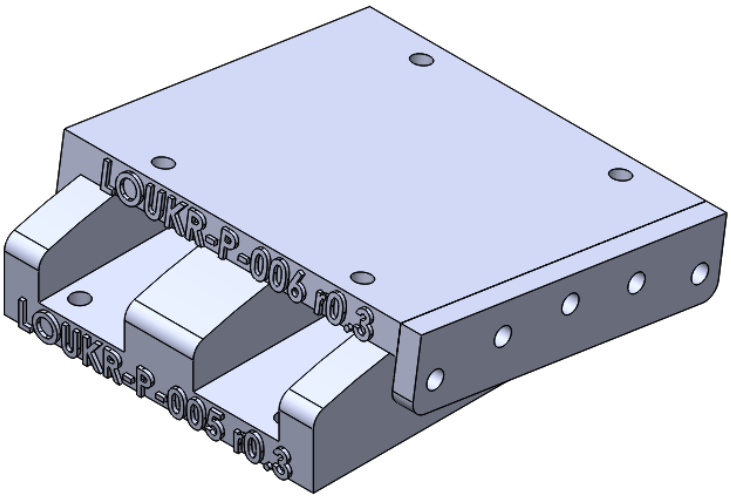


Figure 3.8. Rotating 75 mm arc stage platform with top and bottom halves.

The final design of the mechanical rig is shown in Figure 3.9.

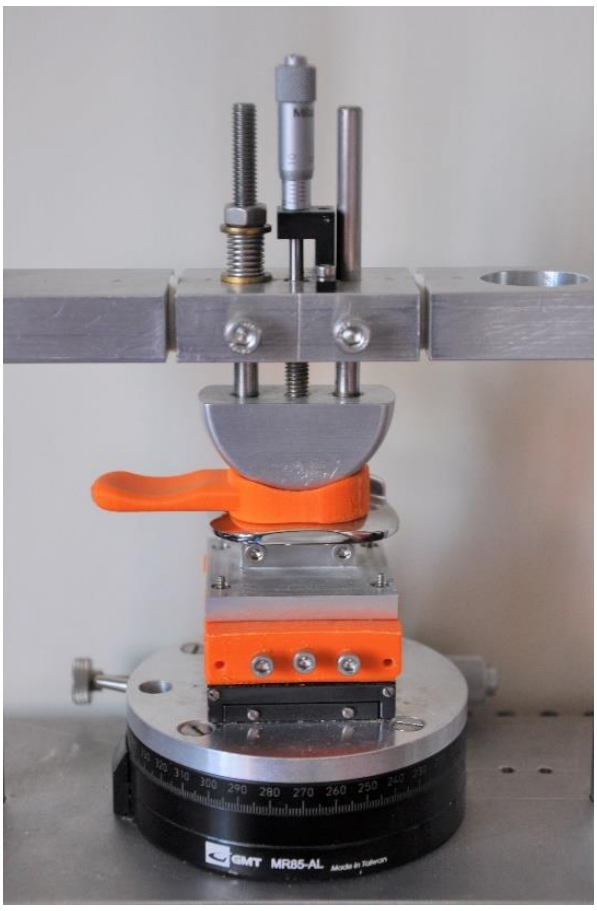


Figure 3.9. Final setup of the mechanical rig for testing. Note the spring in compression on the left hand side. The modified bearing with a handle attached is shown in the picture.

3.3.3.2 *Validating the modified rig results with Western-Simons et al.*

Using the dislocation definition and dislocation manoeuvre described by Weston-Simons et al. [10], it was possible to replicate the medial dislocation results previously reported if the bearings dislocated such that the entire bearing could be removed in the medial direction. This meant that the lateral side of the bearing (the side of the bearing furthest from the tibial wall) cleared both the minimum distance of the femoral component to tibial wall ("a" in Figure 3.7) and the minimum distance of the femoral component to the domed tibial surface ("b" in Figure 3.7). For ML 0 mm – 8 mm, the medial DD was 6.40 mm – 5.00 mm for Weston-Simons et al. [10] and 6.25 mm – 5.00 mm in this study (Table 3.5).

However, the lateral and anterior dislocation results reported in this study were different to those previously reported by Weston-Simons et al. [10]. For lateral and anterior dislocation in this study, the bearing was kept straight/parallel to the ground and to dislocate the bearing, it was pulled straight out either anteriorly or laterally. Lateral and anterior dislocation in this study required 4.00 mm and 7.00 mm DD respectively, compared with 5.00 mm and 8.00 mm in Weston-Simons et al.'s study (Table 3.5).

Table 3.5. Validation testing results comparing the results from Weston-Simons et al. using the original rig with the dislocation results measured in this study using the modified rig.

	Distraction to Dislocation (mm)					
	Weston-Simons et al.'s study [10] using the original rig*			**Current study using the modified rig**		
Distance from the wall (mm)	**Lateral**	**Medial**	**Anterior**	**Lateral**	**Medial**	**Anterior**
0	5.10	6.40	8.20	4.00	6.25	7.00
1	5.10	5.90	8.10	4.00	6.00	7.00
2	5.00	5.50	8.00	4.00	5.50	7.00
3	5.10	5.80	8.00	4.00	5.75	7.00
4	5.10	5.40	8.00	4.00	5.50	7.00
5	5.00	5.40	8.10	4.00	5.75	7.00
6	5.10	5.30	8.10	4.00	5.75	7.00
7	5.10	5.40	8.00	3.75	5.75	7.00
8	4.90	5.00	7.88	3.75	5.00	7.00

*NOTE: the standard deviations were disregarded.

3.3.4 Discussion

As the rig was initially found to be unstable, producing unreliable DD results, Part A of Chapter 3 describes how the linear potentiometer was replaced with a micrometer to enable the rig to record stable and therefore accurate measurements of distraction distances. Plastic sleeves were also added to the crossbar to better constrain vertical distraction motion of the crossbar. Additionally, the rig was also successfully modified to mimic the relative position of the femoral and tibial components when the knee flexes.

Using the modified mechanical rig and the medial dislocation manoeuvre previously described by Weston-Simons et al. [10], it was possible to replicate the medial dislocation results only if it was possible to remove the entire bearing medially, which would require the lateral side of the bearing to

clear both the minimum distance of the femoral component to tibial wall ("a" in Figure 3.7) and the minimum distance of the femoral component to the domed tibial surface ("b" in Figure 3.7). For ML between 0 mm – 8 mm, the medial DD was 6.40 mm – 5.00 mm in Weston-Simons et al. [10] and 6.25 mm – 5.00 mm in this study. For anterior and lateral dislocations, during testing it was noted that dislocations were possible either by keeping the bearing straight/parallel to the ground and pulling it straight out either anteriorly or laterally, or, by keeping the bearing congruent with either the femoral or the tibial surfaces. To achieve congruency with either the femoral and tibial components, the bearing would need to tilt away from the horizontal, which would reduce the DD readings. Despite keeping the bearing straight/parallel to the ground and pulling it straight out either anteriorly or laterally during testing, the DD readings from Weston-Simons et al. [10] were higher than the readings found in this study. For anterior and lateral dislocations, Weston-Simons et al. [10] found DD values of 8.00 mm and 5.00 mm DD respectively, while this study found DD values of 7.00 mm and 4.00 mm respectively.

The most complex dislocation direction was medial dislocation, probably because medial dislocation is obstructed by the tibial wall. During testing it was noted that in addition to the medial dislocation manoeuvre to dislocation described by Weston-Simons et al. [10], it was also possible to dislocate the bearings medially using the herein defined "swing dislocation manoeuvre" in which the bearing was brought into contact with the femoral component so that it was congruent with the femoral component, and subsequently swinging the bearing medially over the wall. The testing was repeated, however, this time using the swing dislocation manoeuvre.

Table 3.6. Repeating Weston-Simons et al. [10] using swing dislocation manoeuvre.

	Medial Distraction to Dislocation (mm)	
Distance from the wall	**Weston-Simons et al. [10] results**	**Swing dislocation manoeuvre**
0	6.40	6.50
1	5.90	6.00
2	5.50	5.50
3	5.80	4.75
4	5.40	4.25
5	5.40	3.50
6	5.30	3.50
7	5.40	3.50
8	5.00	3.50

Repeating the testing using the definition for medial dislocation used by Weston-Simons et al. [10] and the "swing dislocation manoeuvre" (

Table 3.6), it was found that the initial DD values for medial dislocation were very similar for ML distances between 0 – 3 mm, however, beyond 3 mm ML distance, the results began to deviate. At 8 mm ML distance, Weston-Simons et al. [10] reported a DD of 5.00 mm, whereas using the swing dislocation manoeuvre, this study found that just 3.50 mm of DD was required.

Therefore, in Weston-Simons et al.'s study [10], the DD for medial dislocation was probably measured when the bearing had escaped medially, with the entire bearing coming out from between the femoral and tibial components in the medial direction, using a specific manoeuvre to achieve medial dislocation [10]. In the specific manoeuvre the bearing is "*... translated in an anterior direction and extended relative to the femoral component, maintaining full congruence with the femoral component. Secondly, it was internally rotated into valgus until the antero-medial corner was on top of the wall. Finally it was externally rotated and translated medially until the medial edge of the bearing was on top of the wall*". In this study, it was found that the easiest manoeuvre for the bearing to dislocate medially depended on the distance that the bearing was from the tibial wall and the pathway to dislocation used: when the bearing was either flush against the tibial wall (0 mm ML) or 1 mm away from the tibial wall, it was easier to dislocate the bearing medially using the medial dislocation manoeuvre described by Weston-Simons et al. [10]; whereas using the swing dislocation manoeuvre required an extra 0.25 mm of vertical distraction for the bearing to dislocate medially. If the bearing was less than 1 mm from the tibial wall, either dislocating the bearing using the medial dislocation manoeuvre described by Weston-Simons et al. [10] or using the swing dislocation manoeuvre required the same amount of vertical distraction. Based off these results, it was clear, that the term "dislocation" is often used loosely to describe a bearing that is no longer positioned such that it is sitting flat on the tibial surface and congruent with the femoral component.

For lateral and anterior dislocations, it is not clear how these dislocations were measured however, Weston-Simons et al. [10] noted that dislocations were achieved by "*translating the bearing in that direction and rotating the bearing so it maintained fully congruous contact with the femur*", which is similar to the swing dislocation manoeuvre described. However, in this study, replicating the swing dislocation manoeuvre for anterior/lateral dislocations did not allow replications of the results. If the bearing was kept horizontal and pulled straight out in the respective direction of dislocation, it was

also not possible to replicate the results. It is possible that either the bearing used in this study was different to the one described in the study by Weston-Simons et al. [10], or that changes to the rig had affected the results from this study, or even that the handle attached to bearings may also have interfered with the testing in either study, however, the near perfect correlation in medial dislocation results suggest that this is not the case. The exact reason(s) why the lateral and anterior dislocation results reported in this study are not consistent with the results previously reported by Weston-Simons et al. [10] are not clear, however, the offset between the studies was consistent suggesting that the difference in DD values was likely a result of different definitions for dislocation, or due to using a different sequence of movements to achieve dislocation. In Weston-Simons et al.'s study [10], it is not clear exactly what constituted an anterior or lateral dislocation (aside from the fact that the bearings came out in these directions) or how anterior and lateral dislocations were achieved.

3.3.5 Conclusion

In Part A of Chapter 3, the design of the mechanical rig first described by Weston-Simons et al. [10] was also improved to enable the rig to record accurate and reliable DD readings. The rig was also successfully modified to mimic the relative position of the femoral and tibial components when the knee flexes. This study found that for all dislocation directions, more than one pathway existed for the bearing to dislocate. Using the modified rig, it was possible to replicate the medial dislocation results previously reported by Weston-Simons et al. [10] only if the medial dislocation manoeuvre previously described by Weston-Simons et al. was used, and if dislocations occurred such that the entire bearing escaped medially. However, the lateral and anterior dislocation results reported in this study are not consistent with the results previously reported by Weston-Simons et al. [10]; the difference probably is due to the use of different definitions for dislocation and/or different dislocation manoeuvres used to achieve dislocation. Additionally, since using a different dislocation manoeuvre to achieve medial dislocation resulted in different DD values and that the lateral and anterior dislocation values between the studies shared the same trend, it is clear that the lack of clearly defined definitions for dislocations could lead to significantly different DD results. Therefore, it can be concluded that the mechanical rig is a useful tool to allow researchers to quantify the risk of

bearing dislocation, however, clear definitions for dislocation and testing constraints (e.g. manoeuvre to achieve dislocation) are required to ensure consistent and reliable DD readings.

3.4 Chapter 3 Part B: The effect of component position on the risk of bearing dislocation

3.4.1 Aims

In Part B, the first aim was to develop a clear set of definitions for dislocation. Using the definitions for dislocation and the modified rig from Part A, the second aim was to determine the minimum distraction required for dislocation in all directions and to validate these results with another independent user.

3.4.2 Methods

3.4.2.1 *Defining "dislocation"*

To facilitate discussion, a set of clear definitions for the term "dislocation" in the context of mobile bearing UKRs was developed.

3.4.2.2 *Testing dislocations using the modified mechanical rig*

Table 3.7. New dislocation testing parameters.

Testing parameter	Testing range (increments)/values
Bearing overhang values	-6.5 mm†, 1 mm, 10 mm, 17 mm
Rotational angles	-30°*, -10°*, 0°, 10°, 30°
Mediolateral translation (increment)	0.00 – 6.00 mm (0.25 mm increments)
Vertical distraction (increment)	0.00 – 8.00 mm (0.25 mm increments)

†Negative sign indicates underhang

*Negative sign indicates external rotation

The final testing parameters used for the new dislocation assessments are shown in Table 3.7: testing assessed a total of 16 500 implant positions for dislocation. Since in Part A, the rig had been modified and new definitions for testing developed in Part B, it was decided to conduct testing to the same accuracy of 0.25 mm for these results. All data were reported to two decimal places to allow comparison. To assess dislocation, the mobile bearing was manually manipulated using the handle attached to the modified mobile bearing. Dislocations were assessed in four dislocation directions using the mechanical rig: medially over the tibial wall, anteriorly, posteriorly or laterally. To

dislocate the bearings medially, the bearing was allowed to dislocate via any sequence of movements to achieve medial dislocation. For each direction, if more than one sequence of movements resulted in bearing dislocation, the sequence that required the lowest DD was recorded.

The "neutral alignment position" was marked in all graphs as a reference point, showing the point at which the lowest point of the femoral component was vertically aligned with the highest point of the domed tibial surface – see "b" in Figure 3.7.

3.4.2.3 *Validating the results with an independent user: Inter-observer assessment using Inter-class Correlation Coefficient (ICC)*

To validate the dislocation results obtained using the modified mechanical rig, the internal consistency in the medial dislocation data across two independent assessors, myself (IY) and Greta Agustoni (GA) were assessed. GA independently repeated medial dislocation assessment using the modified mechanical rig, testing parameters listed in Table 3.8 and the definitions for dislocation developed under *Section 3.4.3.1 Defining "dislocation"*.

Table 3.8. ICC Validation testing parameters.

Testing parameter	**Testing range (increments)/values**
Bearing overhang values	Implant is central (no bearing overhang)
Rotational angles	0°
Mediolateral translation (increment)	0.00 – 4.00 mm (0.25 mm increments)
Vertical distraction (increment)	0.00 – 8.00 mm (0.25 mm increments)

Testing methodology mimicked that used by IY described under *Section 3.4.2.2 Testing dislocations using the modified mechanical rig*.

For each ML distance, the amount of vertical distraction required for medial, lateral, anterior and posterior dislocation as assessed by IY or GA were then statistically compared using an ICC: two-way random- effects model with type consistency and 95% Confidence Interval (CI) using IBM® SPSS Statistical package (version 25) (SPSS Inc, Chicago, IL).

3.4.3 Results

3.4.3.1 *Defining "dislocation"*

The term "medial dislocation" has always been used to encompass all scenarios in which the medial side of the bearing (the side of the bearing closest to the tibial wall) clears the minimum distance to the tibial wall ("a" in Figure 3.7). In this section, medial dislocation is further divided into three categories called "dislocation levels": level 1, level 2 and level 3. Medial dislocation categories varied depending on how much of the mobile bearing was trapped between the metal femoral and tibial components. Lateral, anterior and posterior dislocations were each also further categorised into three dislocation categories, however, these were called "congruency types", either femoral congruency, tibial congruency or non-congruent, as they differed based on the surfaces with which they were congruent. When testing for dislocations, for each dislocation direction, any category of dislocation was allowed. If for any one configuration more than one dislocation category was possible, the dislocation category that required the least DD was recorded. Dislocation occurrence, dislocation direction and the dislocation category were recorded in a spreadsheet in Excel (version 2016, Microsoft, Redmond, Washington, U.S.).

Medial dislocation

The three levels of medial dislocation (Figure 3.10) were defined based on how much of the mobile bearing was trapped between the metal femoral and tibial components. Level 1 medial dislocation is effectively a fixed subluxation in which the medial edge of the bearing dislocates above the tibial wall, but the lateral edge (the side of the bearing furthest from the tibial wall) does not clear the minimum distance from the femoral component to the domed surface ("b" in Figure 3.7). This meant that if the bearing was not held in that position, the bearing would likely return to a non-dislocated position unless the femoral component drops down while the bearing is dislocated and locks the bearing in this position. Level 2 dislocation occurred when the medial edge of the bearing dislocated above the tibial wall and the lateral edge (the side of the bearing furthest from the tibial wall) cleared the minimum distance from the femoral component to the domed surface ("b" in Figure 3.7) but not the minimum distance from the femoral component to the tibial wall ("a" in Figure 3.7), so it cannot

come out completely medially. Level 3 medial dislocation occurred when the bearing cleared both the minimum distance from the femoral component to the domed surface ("b" in Figure 3.7) and the minimum distance from the femoral component to the tibial wall ("a" in Figure 3.7) and comes completely out medially. If given sufficient vertical distraction, all bearings would eventually reach a level 3 medial dislocation.

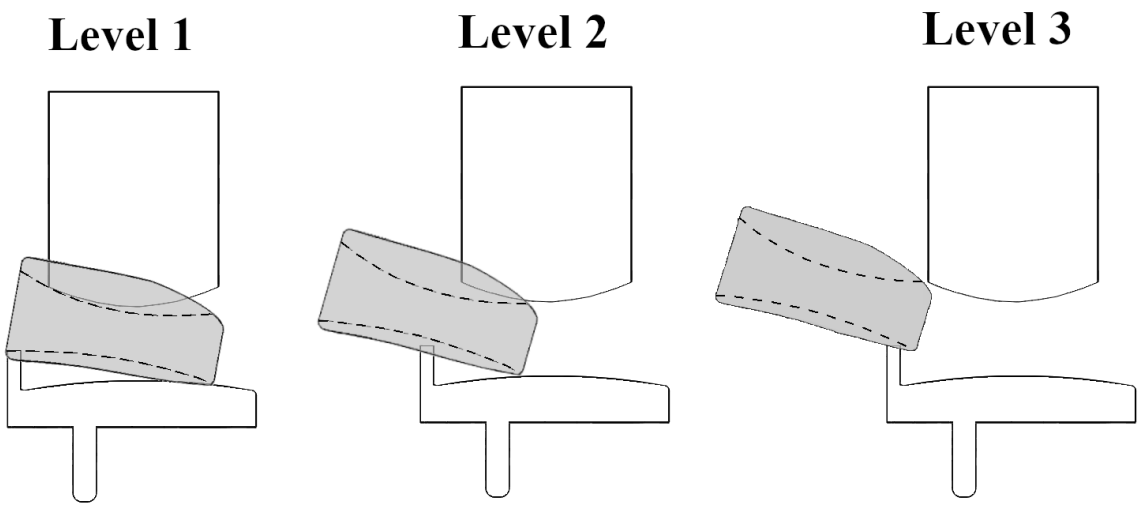


Figure 3.10. Schematic representing level 1, level 2 and level 3 medial dislocation with the Oxford Domed Lateral implant.

Lateral, anterior and posterior dislocation were defined similarly to medial dislocation with three categories of dislocation outlined (Figure 3.11). Femoral congruency dislocations were either anterior, posterior or lateral dislocations which occurred while the bearing remained congruent with the femoral component throughout motion. Tibial congruency dislocations occurred when the bearing remained congruent with the tibial surface throughout motion and non-congruent dislocations occurred when the bearing was not congruent with any surface, but instead, where the bearing remained straight/parallel to the ground and pulled straight out until the bearing dislocated either anteriorly/posteriorly or laterally.

Anterior, posterior and lateral dislocation

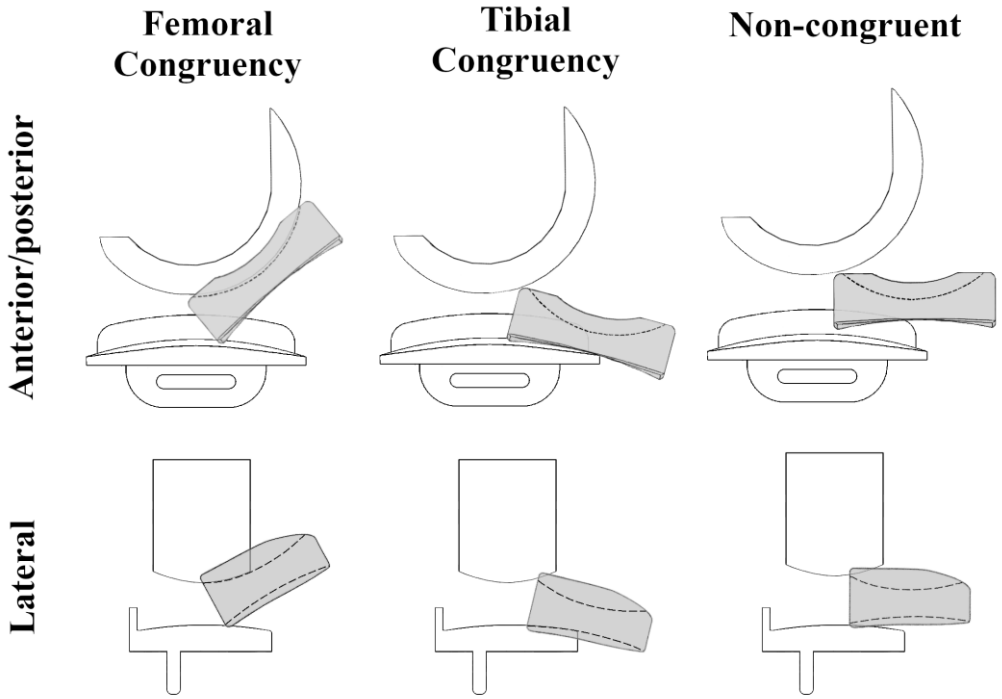


Figure 3.11. Schematic representing femoral congruency, tibial congruency and non-congruent dislocations for lateral and anterior/posterior dislocation with the Oxford Domed Lateral implant.

3.4.3.2 *Testing dislocations using the modified rig*

Table 3.9. The minimum DD (mm) required for medial, lateral, anterior or posterior dislocation of the mobile bearing given various configurations of the metal femoral and tibial components.

		Distraction to Dislocation (DD) (mm)							
		0.00 mm ML distance				6.00 mm ML distance			
Bearing overhang (mm)	**Rotation angle (°)**	**Medial**	**Lateral**	**Anterior**	**Posterior**	**Medial**	**Lateral**	**Anterior**	**Posterior**
-6.5†	**0**	5.50	2.75	5.75	6.00	2.50	2.75	5.75	6.00
	10	5.50	2.75	5.75	6.00	2.25	2.75	5.75	6.00
	-10*	5.50	2.75	5.75	6.00	2.25	2.75	5.75	6.00
	30	5.75	2.75	5.75	6.00	2.25	2.75	5.75	6.00
	-30*	5.50	2.75	5.75	6.00	2.25	2.75	5.75	6.00
1	**0**	5.50	2.75	5.75	6.00	2.25	2.75	5.75	6.00
	10	5.50	2.75	5.75	6.00	2.25	2.75	5.75	6.00
	-10*	5.50	2.75	5.75	6.00	2.00	2.75	5.75	6.00
	30	5.50	2.75	5.75	6.00	2.25	2.75	5.75	6.00
	-30*	5.50	2.75	5.75	6.00	2.00	2.75	5.75	6.00
10	**0**	5.25	2.75	5.75	6.00	2.00	2.75	5.75	6.00
	10	5.50	2.75	5.75	6.00	2.00	2.75	5.75	6.00
	-10*	5.50	2.75	5.75	6.00	2.00	2.75	5.75	6.00
	30	5.50	2.75	5.75	6.00	2.00	2.75	5.75	6.00
	-30*	5.50	2.75	5.75	6.00	2.00	2.75	5.75	6.00
17	**0**	4.50	3.00	5.75	6.00	0.75	2.75	5.75	6.00
	10	4.25	3.00	5.75	6.00	0.75	3.00	5.75	6.00
	-10*	4.25	3.00	6.00	6.00	1.00	3.00	6.00	6.00
	30	4.00	3.00	5.75	6.00	0.75	3.00	5.75	6.00
	-30*	4.25	3.00	6.00	6.00	0.75	3.00	6.00	6.00

†Negative sign indicates underhang

*Negative sign indicates external rotation

Dislocation testing results are summarised in Table 3.9. For all configurations, vertical distraction was required for mobile bearing dislocation. With the metal components at the starting position (bearing flush against the tibial wall, bearing central and without rotation), the amount of DD differed depending on the mode of dislocation and the ML distance. With 0.00 mm ML distance, the DD was 5.50 mm medially, 2.75 mm laterally, 5.75 mm anteriorly and 6.00 mm posteriorly. Increasing the relative distance between the metal femoral and tibial components mediolaterally decreased the DD for medial dislocation to 2.25 mm with 6.00 mm ML distance, however, no effect was seen on the DD for lateral, anterior or posterior dislocations.

Altering the tibial rotation angle (from starting position with 0° tibial rotation to internal/external rotation of 10° or 30°) had negligible effect on the dislocation results, irrespective of the AP position of the bearing (Figure 3.12). Increasing the amount of bearing overhang from central/6.50 mm underhang to 1 mm or 10 mm posterior bearing overhang), also had negligible effect on the dislocation results, irrespective of the rotation angle (Figure 3.13). However, with 17 mm of posterior bearing overhang, the DD was reduced to 4.50 mm for a medial dislocation with 0 mm ML distance and 0.75 mm when the ML distance was increased to 6.00 mm (Figure 3.13). The DD remained 3.00 mm for a lateral dislocation and 5.75 mm for anterior and 6.00 mm for posterior dislocation (Table 3.9).

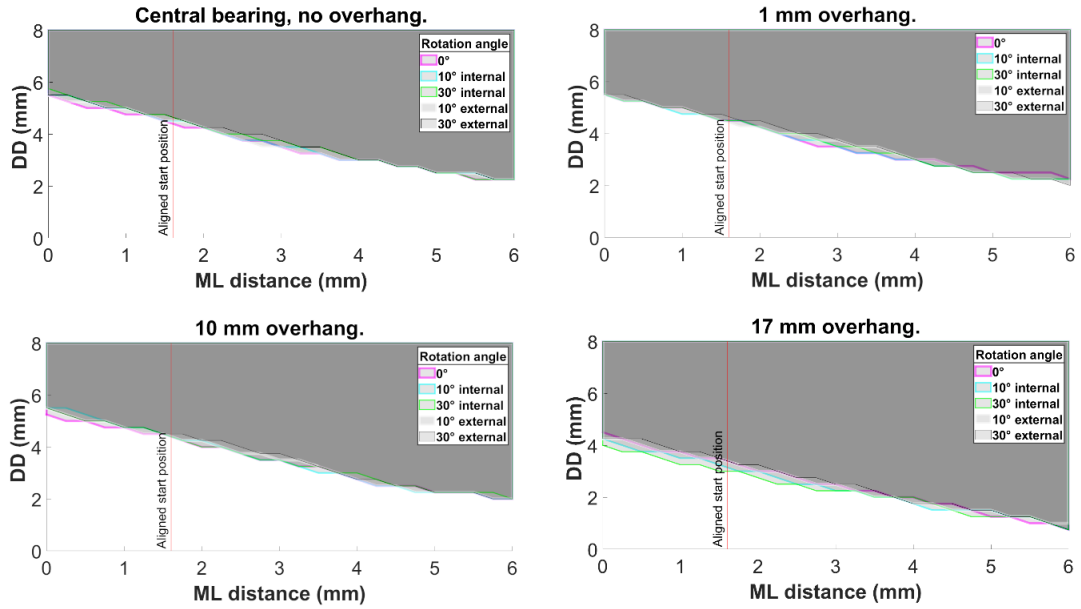


Figure 3.12. The effect of tibial rotation on medial dislocation: Central bearing with no overhang, 1 mm, 10 mm and 17 mm posterior bearing overhang.

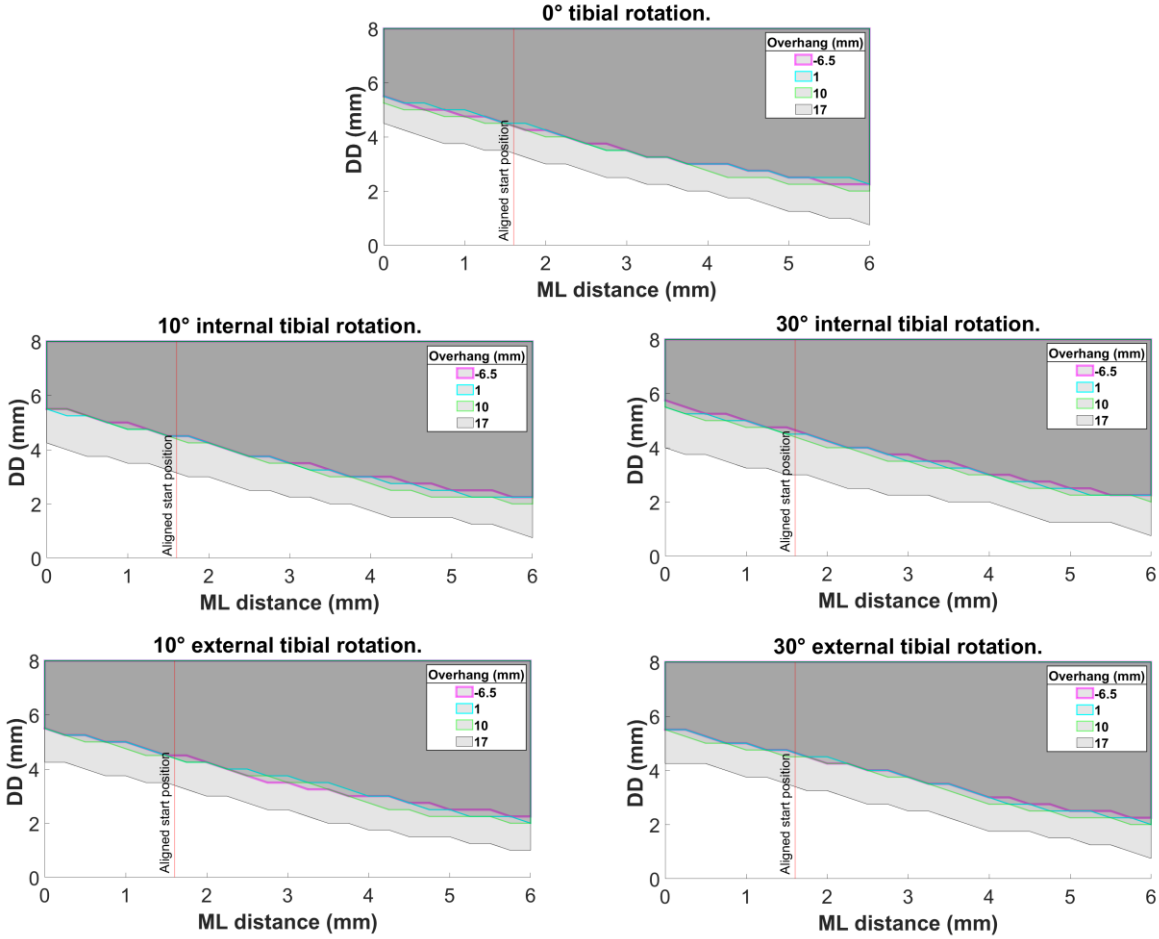


Figure 3.13. The effect of posterior bearing overhang on medial dislocation at 0°, 10° internal, 30° internal, 10° external and 30° external tibial rotation angles.

3.4.3.3 *Validating the results with an independent user: Inter-observer assessment using Inter-class Correlation Coefficient (ICC)*

When comparing medial dislocation results between IY and GA (Figure 3.14 and Table 3.10), for ML 0.00 mm – 4.00 mm, the medial DD was 5.50 mm – 3.00 mm for both studies. The statistical analysis showed excellent inter-observer reliability with an interclass correlation value of 0.995 (95%CI: 0.995 – 0.987) based on a mean rating (k=2), consistency type assessment and two–way random effects model. Lateral dislocation showed complete agreement with a DD of 2.75 mm (Figure 3.14). Anterior/posterior dislocation showed a consistent offset of 1 mm (IY = 5.75 mm and GA = 6.75 mm) (Figure 3.14).

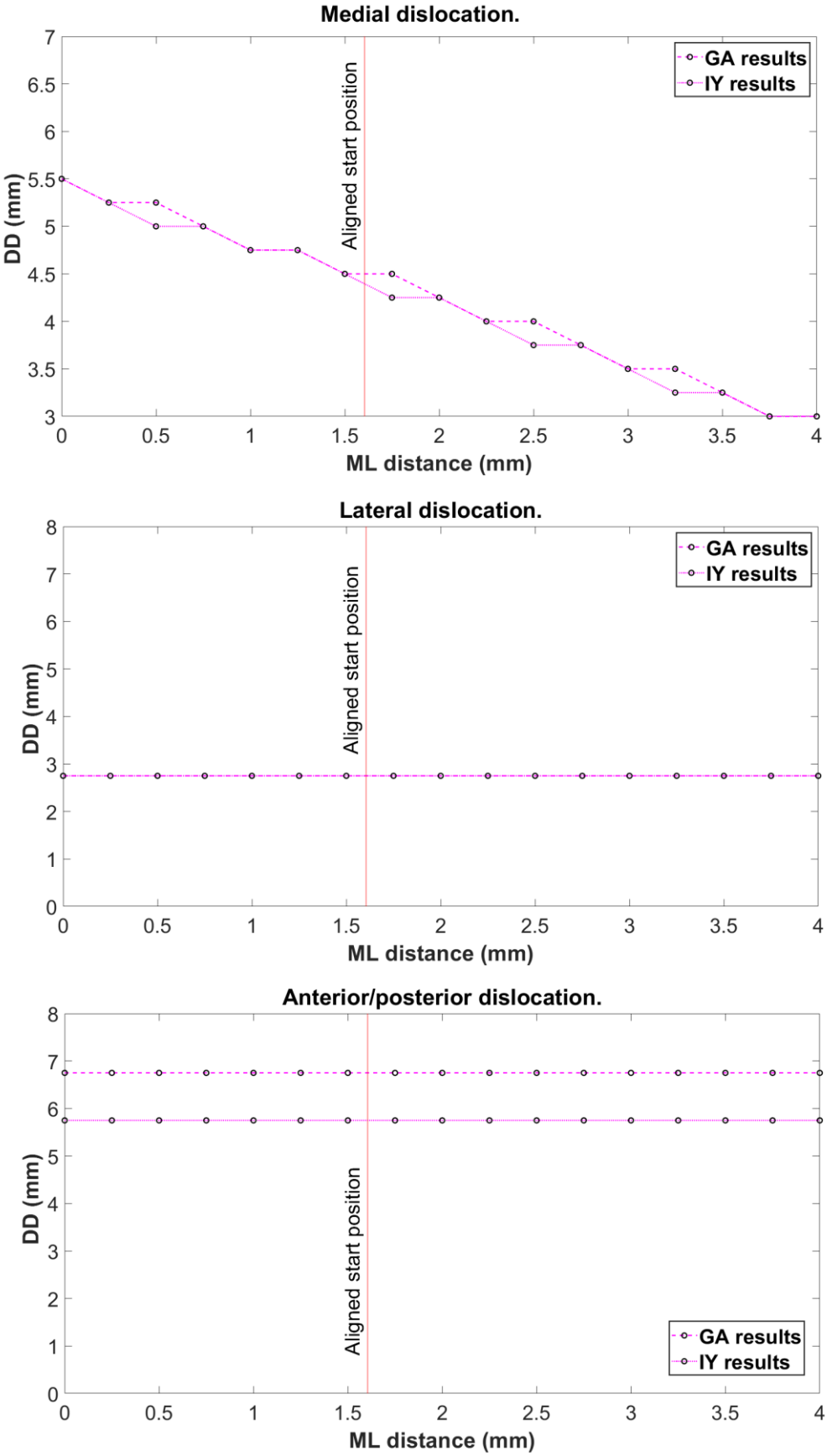


Figure 3.14. Mechanical rig results comparison between IY and GA testing: medial dislocation, lateral dislocation and anterior/posterior dislocation.

Table 3.10. GA results: amount of DD (mm) required for medial dislocation of the mobile bearing given various configurations of the metal femoral and tibial components.

Distraction to Dislocation (DD) Medially (mm)									
		0 mm ML distance				4 mm ML distance			
Bearing overhang (mm)	**Rotation angle (°)**	**Medial**	**Lateral**	**Anterior**	**Posterior**	**Medial**	**Lateral**	**Anterior**	**Posterior**
0	**0**	5.50	2.75	6.75	6.75	3.00	2.75	6.75	6.75

Validation of the results from this study with an independent assessor (GA) confirmed that given a stable mechanical rig and clear definitions for dislocation and the allowed sequence of movements to achieve dislocation, reliable and consistent results can be achieved using the rig as evident by the complete agreement in the lateral dislocation values and excellent correlation for medial dislocation with an ICC of 0.995 (95%CI: 0.995 – 0.987)). For anterior/posterior dislocation, the results between GA and myself differed by 1.00 mm, with IY reporting a DD of 5.75 mm and GA reporting a DD of 6.75 mm.

3.4.4 Discussion

Using the modified rig, the new definitions for dislocation and the dislocation pathway yielding the lowest DD value, the results from Part B represent the minimum DD required for dislocation. For medial dislocations, the minimum dislocation was achieved through level 1 medial dislocation. For lateral and anterior/posterior dislocations it was noted that the minimum dislocation was achieved through femoral congruency rather than tibial congruency, because to make the bearing congruent with the femoral surface required a larger tilt than if the bearing were to be congruent with the tibial surface. The results from validation testing in Part A found anterior dislocations required a DD of 7.00 mm, which was similar to the 6.75 mm reported by GA. On further investigation, it was found that this difference was due to using a different definition for anterior dislocation: GA confirmed that anterior dislocations were achieved using the same technique used by IY in Part A, by keeping the bearing horizontal/parallel with the ground and pulling the bearing out, rather than allowing the bearing to be congruent with the femoral component.

This study showed that distraction of the joint surfaces (measured by the DD) was required for a bearing to dislocate. Further, this study also highlighted how the risk of dislocation varies depending on the direction of dislocation and on the relative ML position of the components, but is otherwise independent of relative component position including internal/external rotation and bearing overhang that can occur with knee flexion angles between 0° and 130°.

Mediolateral translation in this study, like that of Weston-Simons et al. [10], was found to influence the risk of medial dislocation. With the bearing flush against the vertical tibial wall, the DD was 5.50 mm. As the femoral component, and thus the bearing, was moved laterally relative to the tibial component, the DD decreases. At 6 mm ML, the DD is only 2.25 mm, so dislocation would be more likely to occur. As the bearing is furthest from the wall at about 90° flexion [176] this probably explains, in part, why dislocations tend to occur at around this angle. As the surgeon can control the relative ML position of the femoral and tibial components and thus the distance between the bearing and the wall, the bearing should be implanted as close to the wall as possible without hitting it to minimise the risk of dislocation. However, as the Oxford bearing moves on a curved path [10, 157], the bearing needs to be 1 mm or 2 mm from the wall in 90° flexion to prevent the bearing hitting the wall in full flexion or extension.

At present, it is recommended for the femoral component be implanted in a neutral position. There is some evidence suggesting that excessive external rotation of the tibial component should be avoided [177] to prevent bearing dislocation, which Weston-Simons et al.'s study findings supported [10]. However, in this study, it was found that internal and external rotation of 10° or 30° of the tibial component did not change the DD compared to neutral. However, as rotation of the tibial component in the rig was about the centre of the tibial component, the distances that are likely to influence medial dislocation are not changed (i.e. the minimum tibial wall to femoral component, "a", and minimum domed surface to femoral component, "b", in Figure 3.7), it makes sense that these were not associated with a change in DD. However, if this study was repeated but with tibial rotation about the centre of the knee, the results may be different.

This study found that posterior bearing overhang of up to 10 mm, which occurs when the knee flexes between 0° and 130°, had no effect on the risk of mobile bearing dislocation. However in Martin et al.'s study [158] the maximum bearing overhang measured was 17 mm (50% of the bearing), which occurred in a single patient at 140° of knee flexion. With this amount of overhang, the DD medially was reduced (up to 1 mm) whatever the distance between the bearing and the wall. The explanation for this observation is probably that the back corner of the wall is rounded so the apparent height of the wall is about 1 mm lower with 50% bearing overhang than with lower amounts of overhang. Altering the design of the implant so that the corners of the tibial wall are less rounded and instead more square is possible and may assist in reducing the risk of dislocation, particularly when there is a large overhang. However, as 50% overhang was an extreme case in the Martin et al. study [158], it was possibly the result of the tibial component being implanted too far forward. Therefore to avoid this, it is essential that the surgeon ensures that the tibial component is positioned flush with the posterior tibia, however, it is difficult to imagine a situation whereby the lateral compartment of the knee would be distracted at 140° flexion as in this amount of flexion the muscles are likely to be tight, which would limit joint distraction. There is also evidence from an in vivo fluoroscopic study that the maximum laxity of the Lateral Collateral Ligament (LCL) is at 90° of flexion and at higher degrees of flexion it tightens [31]. In addition the descriptions of dislocation have been with the knee flexed but not highly flexed [6, 7, 39, 41, 146, 162]. Taken together this would suggest that extreme overhang at the extreme of flexion is not a major cause of dislocation.

The amount of distraction that occurs in the lateral compartment of the knee depends on many factors such as the position of the knee, the laxity of the ligaments [164] and the loading on the knee [32]. A Magnetic Resonance Imaging (MRI) study found that the lateral compartment of the knee distracted on average 6.7±1.9 mm following application of a varus stress to a knee joint in 90° of knee flexion [32], which is just enough for an anterior or posterior dislocation, whereas it is easily enough for a medial dislocation. As anterior or posterior dislocations are rare, if the surgical technique could be modified so the distraction for a medial dislocation was similar to that for an anterior or posterior dislocation, then the dislocation rate would probably be acceptable. However,

even if the bearing is optimally positioned 2 mm away from the wall, the DD medially is 4.5 mm, which is lower than that for anterior/posterior dislocation, so a different approach is required. It is therefore recommended that during the operation, once bone preparation is complete, trial component should be inserted and the risk of dislocations assessed. If, with the leg in a figure of four position, the trial bearing can easily be dislocated medially a fixed rather than a mobile bearing device should be used. The Fixed Lateral Oxford (FLO) can be used interchangeably with the ODL. Therefore, without changing the bone preparation the FLO should be implanted if there is a concern about medial bearing dislocation.

It is noted that in this study, the final DD reported in Table 3.9 may be higher than reported because between each ML reading the components were not re-zeroed, however the maximum difference is still a very small value of 0.24 mm which occurs when the ML is 6 mm (Figure 3.15).

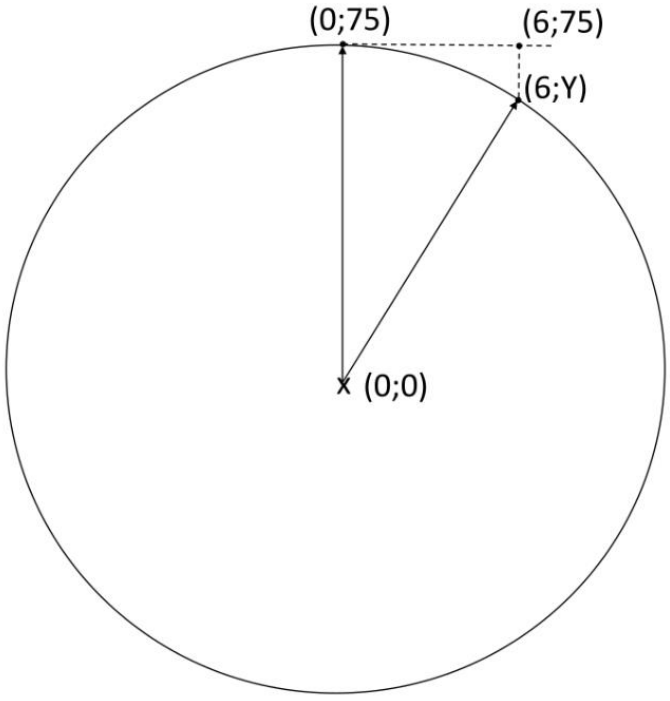


Figure 3.15. Geometric representation of the maximum amount of displacement unaccounted for in testing by not zeroing between readings. Note: this figure is not drawn to scale.

$$Using\ the\ equation\ of\ a\ circle: X^2 + Y^2 = r^2$$

$$6^2 + Y^2 = 75^2$$

$$Y^2 = 6525 - 36$$

$$= 5589$$

$$Y = 74.7596\ mm$$

$$Difference = 75\ mm - 74.7596\ mm$$

$$Difference = 0.24\ mm$$

Despite the lateral edge of the mobile bearing being the same height as the medial edge, lateral dislocations required the least amount of DD (2.75 mm) because medial dislocation is obstructed by the tibial wall. Nevertheless, lateral dislocations tend not to occur clinically probably because the lateral compartment has a curtain of ligaments and other soft tissues laterally which act as a physical barrier preventing lateral dislocations [178]. Interestingly, if surgeons find intra-operatively that they are unable to insert the bearing as usual from the front, surgeons are then instructed to insert the bearing angled at 45° anteriorly. It was found using the rig that 3.00 mm of vertical distraction was required to insert the bearing angled at 45° anteriorly, which is the same amount of distraction required for the bearing to dislocate laterally. This makes sense, since the bearing is angled such that it is essentially accessing the lateral edge to insert the bearing. Further, lateral dislocations require a lateral force applied on the mobile bearing, which is unlikely to occur within the knee. Medial dislocations are more common than anterior/posterior dislocations. One reason for this is probably because the medial DD is smaller than the anterior or posterior DD. Furthermore, for an anterior or posterior dislocation the bearing has to be displaced half its length before it dislocates whereas for a medial dislocation it just needs to sublux a few millimetres onto the wall where it can become entrapped when the lateral compartment is loaded and no longer distracted.

It is clear that a force is required to dislocate the bearing, however, it is not clear what exactly that force is. As the bearing is less dense than the joint fluid [179, 180], the buoyancy force could possibly cause the bearing to just float out, however given that there is only 6.7 ± 2.3 ml (mean ± SD) of synovial fluid in the knee joint [181], this is unlikely the cause of bearing dislocation. Impingement of the bearing on bone, cement, or soft tissue like the popliteus tendon might also cause the bearing

to dislocate [19]. However, whatever is causing the bearing to dislocate, it cannot dislocate unless there is enough distraction in the knee to allow for a dislocation. To minimise the risk of dislocation, surgeons need to do what they can to increase the DD for medial dislocation.

3.4.5 Study limitations

In this study, dislocations were achieved by manually manipulating a modified bearing (handle added to the bearing), which is not reflective of the bearing that would be used in the knee. While we did not find the results in the study to be obviously compromised by the use of a modified bearing, some dislocations e.g. posterior dislocation may have been made more difficult by the handle. Further, while the handle allowed the tester to apply an active force on the bearing to dislocate the bearing, the amount of force applied was not measured so some dislocations may not be possible or realistic. Future work could measure the force used to dislocate the bearing to gain insight into possible forces within the knee joint that could lead to dislocations. While this study concluded that tibial rotation did not impact the risk of bearing dislocation, it is noted that in the rig, rotation of the tibial component was about the centre of the tibial component rather than the centre of the knee. Therefore, if the study were repeated with the centre of rotation located in the centre of the knee, the results may be different, however, clinical studies have found that for optimal functional outcomes following lateral UKR using the ODL implant, the metal femoral and tibial components should ideally be implanted neutrally, without any rotation [10]. Finally, it is noted that testing dislocations with the mechanical rig was done manually, which was both time consuming and subject to user error – variability in results could occur due to tester fatigue or subjective judgement of force applied to achieve dislocation. Using the rig, many factors that might cause the bearing to dislocate once the lateral compartment is distracted, cannot be assessed e.g. the soft tissues in the knee. However, the approach does seem to be valid as it explains why clinically, medial dislocation is most common. In this study, testing was conducted using the most common sized components (medium femoral component, size C tibial component and the 3 mm bearing), however, the results may differ for different component sizes. The effect of different bearing thicknesses was not assessed but this does not seem to influence the dislocation rate [10]. Further, the effect of relative varus/valgus alignment

of the components was also not studied. Further analysis of these parameters and of potential new designs of ODL are required.

3.4.6 Conclusion

The results in the study demonstrated that given a stable mechanical rig and clear definitions for dislocation and the allowed sequence of movements to achieve dislocation, consistent results can be achieved using the mechanical rig. Findings from the study indicate that the reason the ODL bearing dislocates in flexion is not because the components rotate and that the bearing overhangs, but because the lateral compartment of the knee distracts, allowing sufficient room for the bearing to dislocate. Distraction in the lateral knee is possible when the lateral ligaments are lax, which occurs when the knee is flexed, so bearing dislocations are likely to occur when the knee is flexed. The bearings are more likely to dislocate medially than anteriorly or posteriorly because less distraction is needed for a medial dislocation. The further away the bearing is from the wall of the tibial component the greater the risk of dislocation. Therefore surgeons should aim to position the femoral and tibial components so that the bearing is as close to the wall as possible to the tibial wall, but does not jam against it.

3.5 Chapter 3 Discussion and conclusions

In Part A of this chapter, the design of the mechanical rig first described by Weston-Simons et al. [10] was improved to make the rig more mechanically stable. The rig was also subsequently modified to mimic the relative position of the femoral and tibial components when the knee flexes. In Part A, it was found that while it was possible to replicate the medial dislocation results previously reported by Weston-Simons et al. [10], the results for anterior/posterior and lateral dislocations were different. Altering the sequence of movements to achieve medial dislocation resulted in different DD values, highlighting that clearly defining the term "dislocation" is critical to achieving consistent results. Therefore in Part B, clear definitions for dislocation were developed where three categories of dislocation for each dislocation direction (medially, laterally, anteriorly and posteriorly) were described. The bearing was allowed to dislocate via any sequence of movements that would enable dislocation, and the lowest DD value was recorded.

Using the modified rig, the new definitions for dislocation and no restrictions on the sequence of movements to achieve dislocation, the minimal distraction required for dislocation was measured. Validation of these results with an independent assessor confirmed that reliable and consistent results can be achieved using the rig if clear instructions regarding the definition of dislocations and the allowed sequence of movements.

Final testing found that the reason the ODL bearing dislocates in flexion is because when the knee is flexed, the lateral ligaments are lax allowing the lateral knee to distract and therefore, making it more likely for bearings to dislocate. The bearings are more likely to dislocate medially than anteriorly or posteriorly because less distraction is needed for a medial dislocation. The further away the bearing is from the wall of the tibial component the greater the risk of dislocation.

Chapter 4 A COMPUTATIONAL MODEL OF MOBILE BEARING DISLOCATION USING ROBOTICS PATH PLANNING ALGORITHMS

Some of the early results from this chapter have been published in a peer reviewed journal[4], the most up to date findings have been published in a peer reviewed journal[5].

4.1 Introduction

The custom mechanical rig first described by Weston-Simons et al. [10] allows the user to study and quantify the risk of bearing dislocation. Studying dislocation using a modified version of the rig in Chapter 3 was pivotal to understanding the problem of dislocation generally, which led to the development of clear definitions for the term "dislocation" in the context of mobile bearing UKRs and a deep understanding and appreciation of the complex relationship between knee flexion and dislocation.

The ability to quantify the risk of bearing dislocation is useful to compare the dislocation risk of new implant designs with the original ODL design. In other words, the dislocation risk of modified ODL designs aimed at reducing the dislocation could be measured and compared to the current ODL implant to gauge the effectiveness of these changes. However, testing new implant designs in the mechanical rig would require modifications to the rig for each new implant designed, a process which is cumbersome and complex (as demonstrated in Chapter 3 Part A). Furthermore, it also clear from Chapter 3 that there are disadvantages of using the mechanical rig to quantify the dislocation risk: testing with the rig was user dependent and inefficient, and the results obtained using the rig were depended on the definitions used for dislocation.

[4] I. Yang, Jonathan D. Gammell, D. Murray, S. Mellon (2020), Application of a robotics path planning algorithm to assess the risk of mobile bearing dislocation in lateral unicompartmental knee replacement, CAOS 2020. The 20th Annual Meeting of the International Society for Computer Assisted Orthopaedic Surgery, EasyChair series EPiC Series in Health Science, vol 4, pp. 301-5.

[5] I. Yang, Jonathan D. Gammell, D. Murray, S. Mellon (2022), Application of a robotics path planning algorithm to assess the risk of mobile bearing dislocation in lateral unicompartmental knee replacement, Nature Scientific Reports.

Therefore, this chapter presents the results of the development of a computation tool based on the design of the mechanical rig to enable ease of testing new implant designs, improve testing efficiency and to eliminate user dependency.

Initially, a MATLAB based computational tool was developed using 3D point clouds to represent the femoral component and bearing component and a boundary surface to represent the tibial component. The movement of the bearing was controlled mathematically, and collision detection was computed using MATLAB's inShape function to detect when points from the bearing component fell within the tibial component boundary surface. Testing method details and results are presented in Appendix A: MATLAB computational tool to assess bearing dislocation. Compared to the results obtained using the mechanical rig, the MATLAB computational tool was more efficient and successfully eliminated the user. However, using the MATLAB dislocation analysis tool, the dislocation trend for medial dislocation DD was different to the mechanical rig and it was concluded that this was largely due to the highly constrained movement of the mobile bearing. As demonstrated in Chapter 3 and by Weston-Simons et al. [10], for each configuration of the femoral and tibial components of the ODL, if dislocation is possible, there can exist more than one pathway for dislocation, each of which can involve a complex series of incremental steps and orientations that the bearing can take to reach dislocation. While the MATLAB dislocation analysis tool explored many possible orientations for the bearing, it was forced to assess one set of highly defined sequence of movements to achieve dislocation. Therefore, it is possible that some dislocations were missed particularly for more difficult configurations that required either more complex sequence of movements and bearing orientations to achieve dislocation. Defining variations to the sequence of movements to achieve dislocation would require significantly more computational energy than practically feasible, and further assumptions on the possible pathways that the bearing may take to dislocation. Therefore, the research was expanded to explore ways to overcome all of these shortfalls.

The Open Motion Planning Library (OMPL) (Houston, Texas, USA) is an open-source path planning library consisting of various state-of-the-art sampling-based motion planning algorithms [11]. In robotics, path planning algorithms are frequently used to solve challenging problems encountered in

autonomous motion. Rapidly-exploring Random Trees (RRT) [182] is a widely used path planning algorithm. Using RRT, the mobile bearing (the "robot") randomly samples a point in the planning space and assesses whether the bearing can move to that point without colliding with any obstacles in its environment (in this case, the metal femoral and tibial components). If no collision is detected, the robot will move to that new point. This process is repeated, with the bearing moving incremental steps towards the goal, until the centre of mass of the bearing reaches the goal. Using path planning algorithms such as RRT, the bearing can identify a continuous series of collision-free steps connecting the robot's start state to the goal. As more than one pathway can exist from the start to the goal, the algorithm identifies the most efficient route by identifying the path that takes the least steps to reach the goal (Figure 4.1).

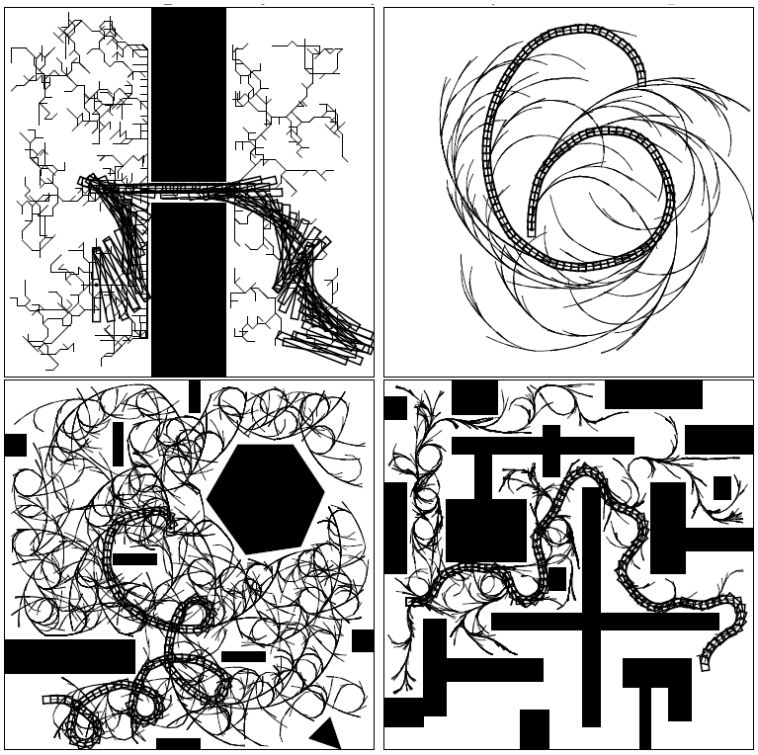


Figure 4.1. Projection of RRT and the optimal computed path for a 2D rigid robot body to navigate in a 2D environment space from a starting to an ending point. Note the obstacles shown in black [182].

4.2 Aims

In this chapter, the aim was to use the robotics path planning algorithm, RRT, and a modified OMPL GUI, to develop a computational tool, termed the "dislocation analysis tool", to measure minimum

amount of vertical distraction of the femoral component relative to the tibial component that is required for a mobile bearing dislocation to occur. In the dislocation analysis tool, the RRT would allow the bearing to independently explore the environment to find a viable pathway to dislocation. The final dislocation results were validated against data obtained using a previously described custom-built mechanical rig (Chapter 3 Part B: The effect of component position on the risk of bearing dislocation).

4.3 Methodology

4.3.1 The setup

CAD models of the ODL components (medium femur, C tibia, and size 3 bearing) were obtained from Zimmer Biomet (Swindon, UK). For the simulation, the OMPL GUI [11] was modified to receive the ODL components: the femoral and tibial components defined the “environment” and the mobile bearing defined the “robot”.

For assessment, a computer with 64-bit Ubuntu 18.04.5 LTS operating system, x64-based processor an Intel Core i7-9850H CPU @ 2.60GHz x 12 processor was used. The installed RAM was 16.0 GB (of which 15.4 GB was usable).

To assess dislocations, a bounding box was defined to limit the search area to the relevant region within the planning space (Figure 4.2). Following initial testing where it was found that the bearing can dislocate anteriorly or posteriorly via the lateral space available, the bounding box for anterior and posterior dislocations was additionally supplemented with a lateral wall, to reproduce the lateral soft tissues that prevent the bearing from moving laterally.

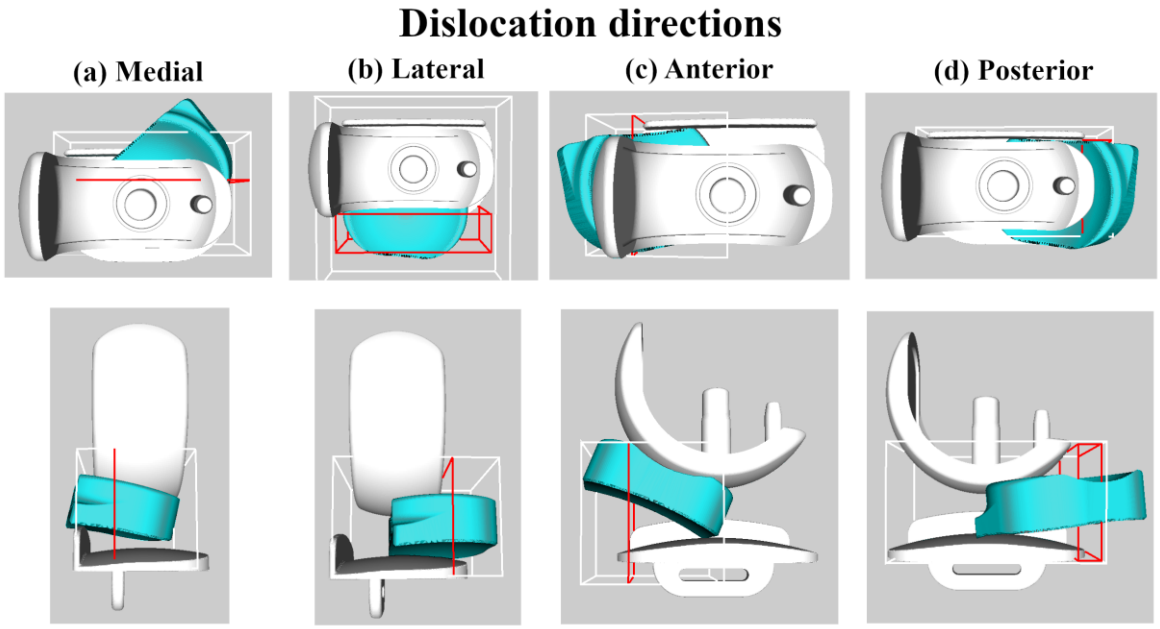


Figure 4.2. Bearing (teal colour) dislocations with bounding box (white outline) and goal area (red outline): (a) medially, (b) laterally, (c) anteriorly or (d) posteriorly. The top panel shows top view and the side view is shown in the bottom panel.

For medial, lateral, anterior and posterior dislocations, a "goal area" was defined which was used to bias the search direction and to identify dislocations. Dislocations occurred when the centre of mass of the mobile bearing entered the "goal area" (Figure 4.2), located within the bounding box.

4.3.2 Implementing the "smart search"

To improve the flexibility of the tool, the initial tool was modified to enable automatic identification of a valid start position which allowed the tool to correct for minor surface overlaps. This feature was particularly useful when, for some configurations, the bearing is translated mediolaterally causing minor overlap of the congruent surfaces between the inferior bearing surface and the domed tibial surface.

The initial brute search algorithm was modified to enable smart searching, whereby for each mediolateral distance, directional searching was implemented to allow the algorithm to begin the search from the configurations most likely to dislocate (maximum vertical distraction), to the configurations least likely to dislocate (minimum vertical distraction), starting with the smallest

mediolateral translation distance. If no dislocations were identified in the allocated initial search time of 180 seconds and 10 search attempts, the configuration was re-searched with an extended time of 270 seconds and 25 search attempts. In other words, as the algorithm approached configurations that were least likely to dislocate (minimal vertical distraction between the femoral and tibial components), the search capacity was increased. If the extended search still failed to identify a dislocation, the algorithm would progress to assessing the next mediolateral translation distance. This was done to improve the likelihood of the tool identifying more difficult dislocations whilst maintaining the efficiency of the search tool.

4.3.3 Selecting search parameters

The RRT parameters (allowable search time & number of search attempts) were tested to select an improved testing time and number of searches. The relative distance between the femoral and tibial components was increased: vertically (0–8 mm) with mediolateral distance set at 2 mm (33 configurations). Based on a process of trial and error, the path step length was set to 1.25 mm as this value resulted in good performance for the dislocation analysis tool and collision resolution was set to 0.0001 mm.

To determine the search time, the number of search attempts was fixed at 10. The time for which the RRT algorithm was applied to the mobile bearing was increased from 15-405 seconds in 15 second increments (26 configurations) to assess whether dislocation was possible, and to record the minimum DD.

To determine the number of searches, the RRT algorithm search time was fixed at 180 seconds. The number of attempts made to search for bearing dislocations was increased from 5-35 searches in 5 search increments to assess whether dislocation was possible, and to record the minimum DD.

4.3.4 Testing

The starting position was defined with the bearing sitting flush against the tibial wall and congruent with the domed tibial surface. The relative distance between the femoral and tibial components was

increased: vertically (2–8 mm) and mediolaterally (ML) (0–6 mm) in 0.25 mm increments (425 configurations). Since testing using the mechanical rig was accurate to ¼ mm, the same accuracy of testing was used for these results to allow ease of comparison. All data were reported to two decimal places. For each configuration, the RRT algorithm was applied to the mobile bearing to assess whether dislocation was possible, and to determine the minimum DD. Path step length was set to 1.25 mm and collision resolution to 0.0001 mm. For each configuration, the RRT algorithm was allowed to run for 270 seconds and 10 searches. If no solution was found in this time, the search was repeated for the same configuration but with the time and search attempts extended to 405 seconds and 25 searches, respectively. Dislocations were labelled according to whether they occurred medially, laterally, anteriorly or posteriorly.

Results obtained using the robotics dislocation analysis tool were validated against results obtained using a previously described mechanical rig (Chapter 3 Part B: The effect of component position on the risk of bearing dislocation). For each mediolateral distraction distance, the amount of vertical distraction required for dislocation was determined using either the mechanical rig or the robotics analysis tool. Results were then statistically compared using an Intraclass Correlation Coefficient (ICC): two- way random- effects model with type consistency and 95% Confidence Interval (CI) using IBM® SPSS Statistical package (version 25) (SPSS Inc, Chicago, IL).

4.4 Results

The final code developed can be found under *Appendix B: Robotics dislocation analysis code*.

4.4.1 Selecting optimal search parameters

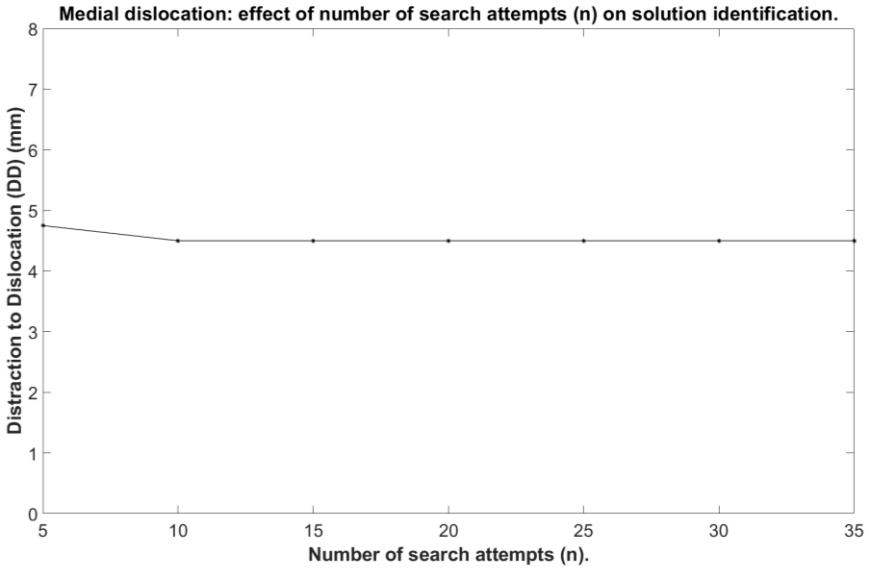


Figure 4.3. The effect of number of searches (n) on solution identification.

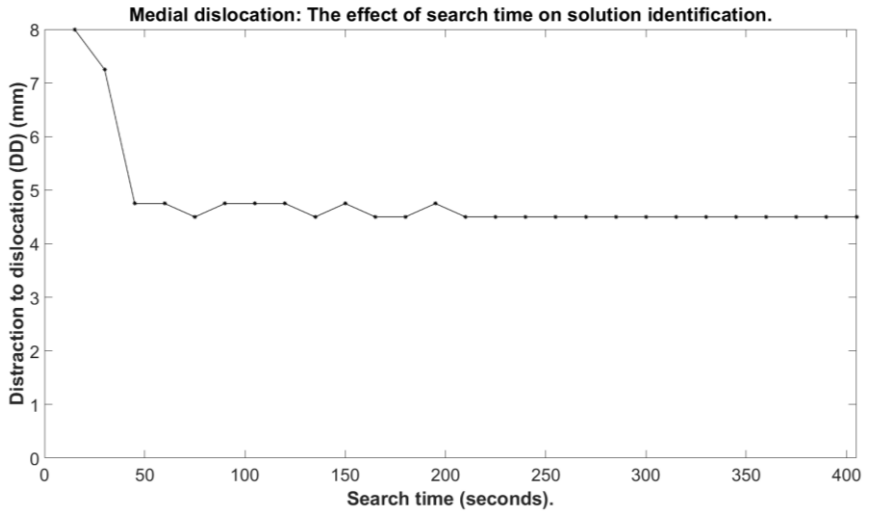


Figure 4.4. The effect of search time (s) on solution identification.

The OMPL GUI was successfully modified to receive the ODL implant components. Preliminary testing identified that at least 10 iterations (Figure 4.3) and 210 seconds (Figure 4.4) are required for maximum solution identification. However, within 10 search attempts (Figure 4.3) and 45 seconds (Figure 4.4) most of the solutions had been identified.

4.4.2 Validation of dislocation results

Table 4.1. Top: Dislocation results obtained using the mechanical rig vs. the robotics dislocation analysis tool.

Distraction to Dislocation (DD) (mm)								
	Mechanical rig				Robotics dislocation analysis tool			
ML translation (mm)/Dislocation mode	Medial	Lateral	Anterior	Posterior	Medial	Lateral	Anterior	Posterior
0.00	5.50	2.75	5.75	6.00	5.50	3.50	6.25	6.25
6.00	2.75	2.75	5.75	6.00	3.25	3.00	6.25	6.00

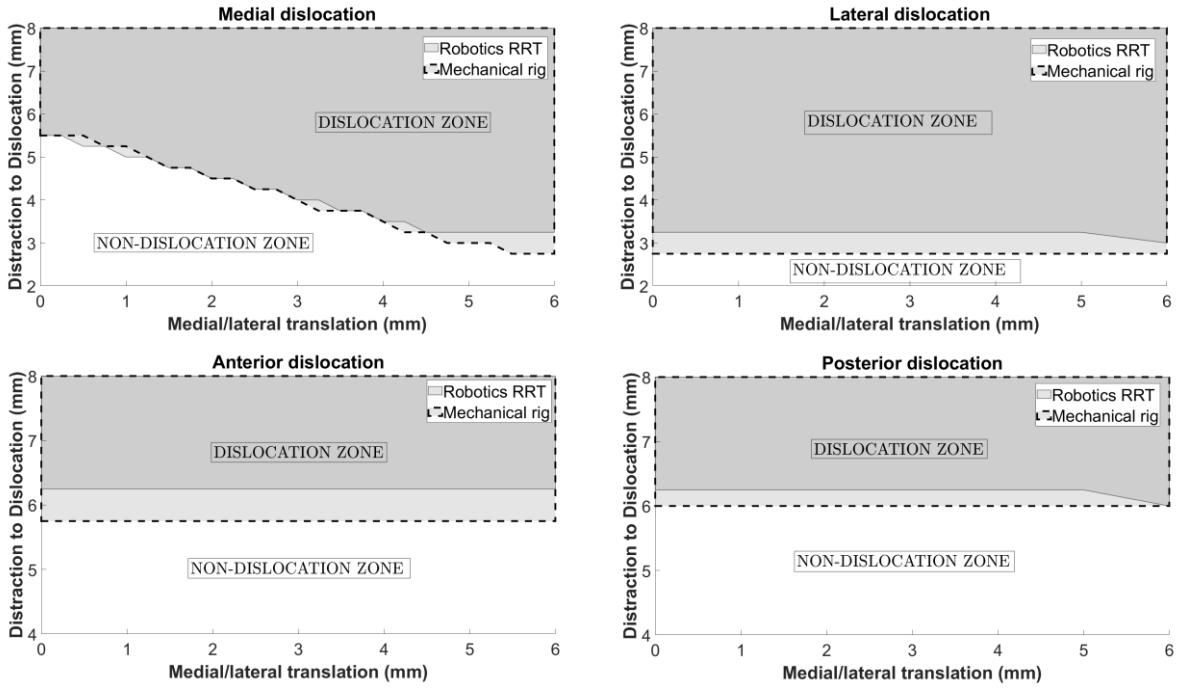


Figure 4.5. Comparing the mobile bearing dislocation (medially, laterally, anteriorly or posteriorly) results obtained using the robotics dislocation analysis tool versus mechanical rig. The grey region marks the dislocation zone and the white region marks the no dislocation zone.

The DD using the dislocation analysis tool was compared with that obtained using the mechanical rig for dislocations medially, laterally, anteriorly and posteriorly (Table 4.1 and Figure 4.5). Anterior, posterior and lateral dislocations were independent of ML position: the robotics dislocation analysis tool output DDs which were marginally higher than the manual mechanical rig: 6.25 mm versus 5.75 mm anteriorly, 6.25 mm versus 6.00 mm posteriorly and 3.25 mm versus 2.75 mm laterally. In contrast, medial dislocation was dependent on the ML position: as ML translation increases from 0 mm to 6 mm, the DD medially of the mobile bearing decreases. For the mechanical rig, the reduction was from 5.50 mm to 2.75 mm, and for the dislocation analysis tool, the reduction is from 5.50 mm

to 3.25 mm (Table 4.1 and Figure 4.5). The results appear to deviate between 4.50 – 6.00 mm of ML translation.

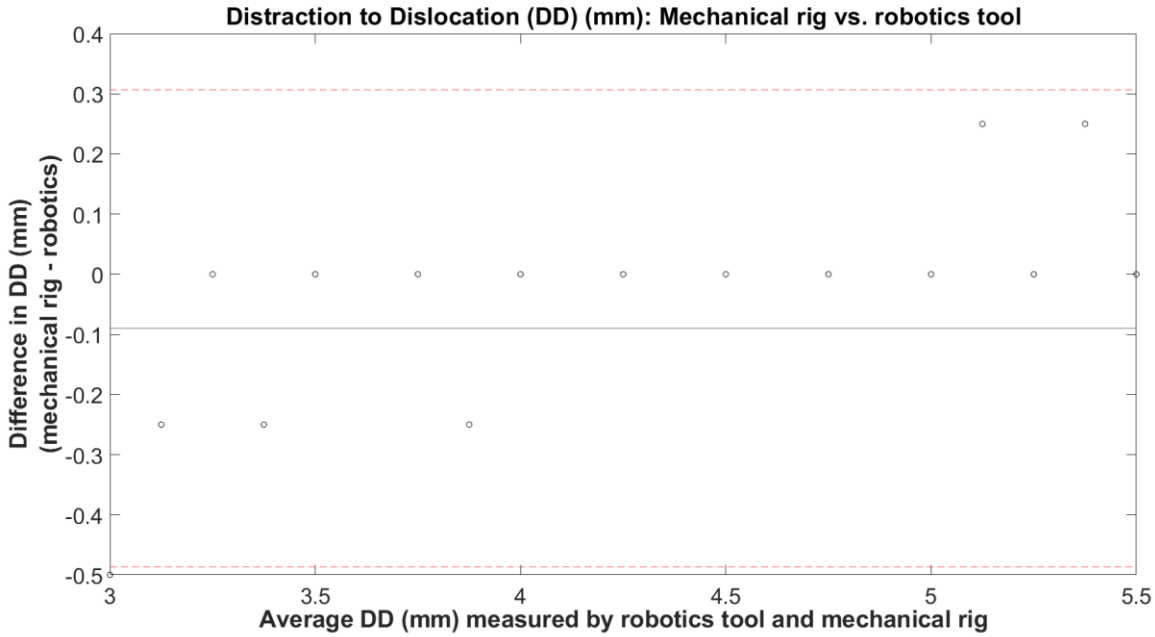


Figure 4.6. Bland-Altmann assessment for medial dislocation comparing mechanical rig vs. robotics dislocation analysis tool.

Compared with mechanical rig, the medial DD when using the dislocation analysis tool differed on average by 0.09 mm (standard deviation: 0.20 mm) (Figure 4.6).

When comparing medial dislocation results between the mechanical rig and the robotics dislocation analysis tool, the statistical analysis showed excellent reliability [183] with an intraclass correlation value of 0.993 (95%CI: 0.982 – 0.998) based on a mean rating (k=1), consistency type assessment and two-way random effects model.

4.5 Discussion

This study demonstrates a novel and successful application of a robotics path planning algorithm, RRT, to the clinical problem of mobile bearing dislocation. The robotics dislocation analysis tool identifies dislocations of the mobile bearing given various configurations of the femoral and tibial

components. It functions by applying the RRT algorithm to the mobile bearing, automating the movement of the mobile bearing from a non-dislocated position to a dislocated position.

The tool was been optimised through implementation of directional searching (starting with the most likely dislocations to the least likely dislocations), with the ability to ramp up search parameters on approaching the configurations for which dislocation is less likely. Finally, the tool was also made more agile, by enabling automatic identification of a valid start state. Path planning algorithms like RRT are probabilistically complete in that the probability of finding a viable path (if one exists) from the start to the goal is one, if the tool is given infinite time and unlimited solve attempts to find the path [184]. However, in this study, not all environment configurations had a viable solution. Therefore to ensure that all configurations that could result in dislocation are identified, the search time and number of searches were incrementally increased when encountering more difficult configurations where dislocations were less likely. All of these modifications, made the tool more stable, and accurate.

For an anterior or posterior bearing dislocation, the DD was between 5.50 mm and 6.50 mm under all circumstances. In contrast, for a medial dislocation the DD was generally substantially less and could be as low as 3.50 mm. It was also influenced by many factors such as the bearing thickness, the ML position of the bearing and the wall height. The lower DD medially than anteriorly or posteriorly explains why medial dislocations occur much more frequently than either anterior or posterior dislocations.

The ML position of the bearing had a marked effect on the DD: the further the bearing is from the wall the lower the DD and thus the higher the risk of dislocation. The surgeon can control the ML position of the bearing and should make it as close to the wall as possible. During rotation and flexion/extension the lateral bearing tends to move on an approximately curved track, around the medial compartment [10], so the bearing is closest to the wall in extension and full flexion. Before making the slot for the keel the surgeon should adjust the ML position of the tibial component so the bearing does not push against the wall in full extension and flexion. This will result in it being about 2 mm or more from the wall at 90° flexion. Given that the position of the vertical saw cut can only

easily be adjusted in 2 mm increments the surgeon should be able to position the tibial component so the bearing is 2 mm to 4 mm from the wall.

This study used the most common path planning algorithm, RRT, for testing. Other path planning algorithms exist [185], which differ in how the problem is defined and how a solution is found [186]. Using a different planner may offer further improvements in the performance of the dislocation analysis tool (e.g. reduce search time and/or number of search attempts), however, since optimising the RRT planner resulted in excellent results, it was not necessary to explore any other planners.

The testing parameters of DD and ML translation were chosen based on physiologically relevant constraints. For the DD, as maximum distraction has been found to be 6.7±1.9 mm [32], testing assessed DD from 0 – 8 mm. Similarly for ML translation, surgeons are unlikely to implant the bearing more than 6 mm apart mediolaterally, so testing assessed 0 – 6 mm.

The dislocation analysis tool was validated by comparing its results to the mechanical rig results. Using the robotics dislocation analysis tool, the DD medially differed by 0.09 mm (standard deviation of 0.20 mm). However, with altering ML translation, the trends were the same suggesting the analysis tool is valid and overall, statistical analysis of medial dislocation showed excellent agreement with an intraclass correlation value of 0.993 (95%CI: 0.982 – 0.998). Reflecting on the results from Weston-Simons et al. [10] where medial DD was found to be for ML between 0 mm – 8 mm, the medial DD was 6.40 mm – 5.00 mm, the results from the robotics dislocation analysis tool confirms that the results found using the modified rig and the new definitions for dislocation as defined in Chapter 3 more closely resemble the effect of implant component positioning on the risk of bearing dislocation.

The discrepancy between the results, particularly between 4.50 - 6.00 mm for medial dislocations, is likely due to experimental error with the mechanical rig, using 3D printed parts, and with user error likely affecting the results (see further details in 3.4.5 Study limitations). With the robotics analysis tool, inherent randomness of RRT may result in false negatives or failure to identify solutions, however, this error was minimised through optimisation of the RRT search parameters: the number of search attempts and the search time. Further, goal sensitivity in the robotics tool may affect more

subtle dislocations where the bearing can lodge on top of the tibial wall, but is unlikely to completely come out from between the femoral and tibial components. Further tweaking can minimise the deviation seen. However, for the purposes of the tool, accuracy of within 0.50 mm is sufficient to satisfy the clinically relevant threshold of 1 mm.

As the laxity of the knee (and thus the maximal possible distraction) varies between patients, the risk of dislocation was assessed by the minimum amount of distraction that would allow a dislocation to occur, given any setup of the femoral and tibial components. The DD medially (3.25 to 5.50 mm) was much less than that for an anterior or posterior dislocation (6.000 or 6.25 mm). This explains why medial dislocation is more likely to occur than anterior or posterior dislocation.

Tokuhara et al. [32] studied the laxity of the lateral side of the knee in flexion and showed that the lateral compartment of the knee distracted on average 6.7±1.9 mm. This is similar to the amount required for an anterior or posterior dislocation, probably explaining why these dislocation modes are rare. The amount of distraction required for lateral dislocation was less than that required for a medial dislocation. However, when the lateral compartment is distracted the lateral ligament and other soft tissues are tight [164], preventing a lateral dislocation.

While this study described the use of the tool using the ODL implant, the tool may be used in the same way with other mobile bearing UKR designs. Further, the tool has other uses beyond finding the DD. Firstly, the manner in which mobile bearings dislocate is not well understood. The tool has enabled us to gain insight for the first time, into how the bearing might be moving in order to reach a dislocated position. Further work to characterise this motion, may enable us to gain a better understanding on the possible causes of dislocations. Secondly, since the tool can assess any setup of the femoral and tibial components, the tool may be used to test the effects of surgical alignment on dislocation of the mobile bearing, or to identify the ideal parameters for surgical implantation of the ODL implant. Finally, the tool can be used to test new ODL implant designs, which may reduce the risk of mobile bearing dislocation.

4.5.1 Study limitations

The dislocation analysis tool allows the study of potential pathways that the bearing may take in order to dislocate. However, the tool cannot take into account many factors, in particular those that might cause the bearing to dislocate once the lateral compartment is distracted (e.g. the ligaments, synovial fluid and the soft tissues in the lateral knee) and so some dislocations identified by the tool may not be possible in a real knee joint. The tool assesses dislocation as a static geometric problem without any forces whereas in reality, the knee is a highly dynamic joint where active forces during knee motion are likely to contribute to driving the bearing into a dislocated position. Future work could aim to design a dynamic dislocation analysis tool which would assess bearing dislocation in a dynamic environment. Despite these limitations, the dislocation analysis tool successfully replicates findings obtained using a mechanical rig and offers a potential explanation for why medial dislocation is the most common mode of dislocation clinically. The dislocations analysis tool is probabilistically complete, which means that if no restrictions are imposed on the search parameters, any pathway to dislocation that exists, will be found. Therefore, to reduce the experimental error, further work would also significantly increase the testing parameters, to ensure that the search time and number of testing attempts, does not artificially eliminate possible dislocation configurations.

4.6 Conclusions

This study describes a novel and validated robotics dislocation analysis tool, which can identify mobile bearing dislocations (medially, laterally, anteriorly or posteriorly) given any setup or design of the femoral and tibial components of the ODL implant. Using the tool, it is possible to measure the DD. Since a smaller DD means that less distraction is required for dislocation to occur, it is thought that dislocations are more likely. Therefore, DD is inversely related to the dislocation risk and can be used to compare the relative dislocation risk of modified ODL designs with respect to the original ODL implant design. Additionally, the tool may be used to identify the ideal parameters for surgical implantation of the implant, or to assess potential new designs that aim to decrease the bearing dislocation rate.

Chapter 5 COMPARISON OF DISLOCATION RISK TO DISLOCATION RATES: FLAT VERSUS DOMED OXFORD IMPLANT

5.1 Introduction

Despite dislocation of the mobile bearing being the largest complication with the ODL implant in the lateral compartment of the knee [6, 122], little is known about the direction of the dislocations since it is not routine practice to report the dislocation direction. As the mobile bearing is unconstrained, it can theoretically dislocate anteriorly, posteriorly, medially or laterally. Clinically, lateral dislocations have not been reported, whereas dislocations anteriorly, posteriorly and medially have been reported using either the Oxford flat lateral implant [90, 122, 163] or the ODL implant [6, 8, 41].

Based on the systematic review results presented in Chapter 2, changing the design of the tibial component from a flat tibial surface to a domed tibial surface reduced the overall clinical dislocation rate [9], however, it is not clear what effect this design change had on the clinically relevant dislocation directions (anterior, posterior or medial).

5.2 Aims

The aim of this chapter was to determine how changing the design of the tibial component from a flat to a domed tibial surface design affected the dislocation risk. Using the dislocation analysis tool presented in Chapter 4, the dislocation risk was measured medially, anteriorly and posteriorly comparing the results when using either the Oxford flat lateral implant or the Oxford Domed Lateral (ODL) implant. A systematic review was also conducted to assess the clinical dislocation rate in each direction for the Oxford flat lateral implant or the Oxford Domed Lateral (ODL) implant. The final portion of this chapter compares the computational dislocation risk results with the systematic review findings for the clinical dislocation rate.

5.3 Methods

5.3.1 Dislocation rates calculated from literature

Using the systematic search strategy used in Chapter 2 (2.2.1 Search strategy), all Randomised Control Trials (RCTs), cohort studies, case control studies, case series studies reporting on the outcome of adult patients treated with either the Oxford flat lateral implant or the ODL implant with details on the dislocation modes were included in the analysis. All reviews were excluded. Studies assessing radiographic imaging or conducted in animals, models (mechanical and computational), and in vitro were excluded from the study. Case reports were also excluded because it was not possible to calculate the dislocation rate. Registry data studies were also excluded due to insufficient data specific to the Oxford lateral implants (either the flat tibial component or the ODL implant).

For each study, where possible, the following information was extracted: first author, publication date, total number of lateral knees operated, study start date, study end date, number of bearing dislocations reported and as best as possible, the mode of dislocation for each dislocation (anterior, posterior, medial, lateral). In this study, since lateral dislocations do not occur clinically, medial dislocations are usually explicitly stated (probably as it is the most common dislocation direction), the remaining dislocations where the direction was not explicitly recorded were considered as either anterior/posterior dislocations.

5.3.2 Dislocation risk calculated computationally

Computer Aided Design (CAD) 3D models of the components of the Oxford flat lateral implant and the ODL implant were modelled using SOLIDWORKS 2021 (Dassault Systemes SOLIDWORKS Corp., Waltham, MA, USA). For the Oxford flat lateral implant, the components modelled were a medium size femoral component, size C flat lateral tibial component and size 3 flat bearing. For the ODL implant, the components modelled were a medium size femoral component, size C domed lateral tibial component and size 3 domed bearing. Stereolithography (STL) files were generated and

used in the robotics dislocation analysis tool to computationally assess the risk of medial, anterior and posterior dislocations.

For both experiments using either the Oxford flat lateral implant and the ODL implant, the testing was conducted using the same testing methodology as in Chapter 4 under *Section 4.3.4 Testing*, except the vertical distraction testing was conducted to an accuracy of 0.25 mm and mediolateral translation accuracy was chosen to an accuracy of 1 mm. All data were reported to two decimal places to allow comparison. The DD results were plotted against each ML distance for the Oxford flat lateral implant and the ODL implant. Separate plots were made to compare the two implants for each mode of dislocation: medial dislocation, anterior dislocation and posterior dislocation.

5.4 Results

5.4.1 Dislocation rates by dislocation direction

Table 5.1. Clinical dislocation rates by dislocation direction: flat tibial component design.

				% dislocation rate		
Study	**Year**	**Start & end date**	**Total no. of lateral knees**	**Overall dislocations**	**Medial dislocations**	**Dislocation direction not specified**
Goodfellow et al. [122]	1988	Nov 82 – Nov 85	27	5/27 (19)	2/5 (40)	3/5 (60)
Pandit et al. [7]	2010	Apr 98 - Oct 04	65	3/65 (5)	2/3 (67)	1/3 (33)
TOTAL	**See dates above**	**See dates above**	**92**	**8/92 (9)**	**4/8 (50)**	**4/8 (50)**

Of the five dislocations that occurred in the lateral knee reported by Goodfellow et al., two were reported as medial dislocations where the bearing subluxed on top of the wall of the tibial component, and for the remaining three dislocations, the mode of dislocation was not explicitly reported [122] so these were recorded as anterior/posterior dislocations. The final series reporting on the flat lateral Oxford was in the phase III implant published by Pandit et al. [7] where three dislocations occurred

in sixty-five knees – two were medial dislocations, while the mode of dislocation for the third dislocation was not mentioned, so it was recorded as anterior/posterior dislocation.

As seen in Table 5.1, combining these two studies, there was a total of ninety-two lateral knees that were operated on using the Oxford flat lateral implant where there was an overall dislocation rate of 9% of which approximately 50% of dislocations occurred medially and the remaining 50% of dislocations occurred either anteriorly or posteriorly.

Table 5.2. Clinical dislocation rates by dislocation mode: domed tibial component design.

				% dislocation rate		
Study	**Year**	**Start & end date**	**Total no. of lateral knees**	**Overall dislocations**	**Medial dislocations**	**Anterior/posterior dislocations**
Schelfaut et al. [8]	2013	Jan 09 – Apr 11	25	1/25 (4)	1/1 (100)	0/1 (0)
Fornell et al. [41]	2018	Jun 08 – May 15	41	1/41 (2)	1/1 (100)	0/1 (0)
Kennedy et al. [6]	2019	Sep 04 – Dec 15	325	14/325 (4)	10/14 (71)	4/14 (29)
TOTAL	**See dates above**	**See dates above**	**391**	**16/391 (4)**	**12/16 (75)**	**4/16 (25)**

For the ODL implant, Schelfaut et al. reported a medial dislocation in a series of twenty-five lateral knees [8]. Similarly, Fornell et al. reported one medial dislocation in a series of forty-one knees [41]. And finally in the largest series reported yet for the Oxford Knee, Kennedy et al. reported fourteen dislocations in 325 lateral knees, of which ten dislocations occurred medially, two dislocations occurred anteriorly and the remaining two dislocations did not have the mode of dislocation reported [6] so these were recorded as anterior/posterior dislocations.

Combining these three studies, there was a total of 391 lateral knees that were operated on using the ODL implant where there was an overall dislocation rate of 4% of which approximately 75% of dislocations occurred medially and 25% of dislocations occurred either anteriorly/posteriorly.

As seen in Chapter 2, changing the design of the tibial component from a flat tibial surface to a domed tibial surface reduced the overall dislocation rate from 17% to 3.7%. Based on these results and if we assume that remaining dislocations are anterior/posterior dislocations, Figure 5.1 shows a flow chart of the inferred clinical dislocation rates separated by dislocation directions.

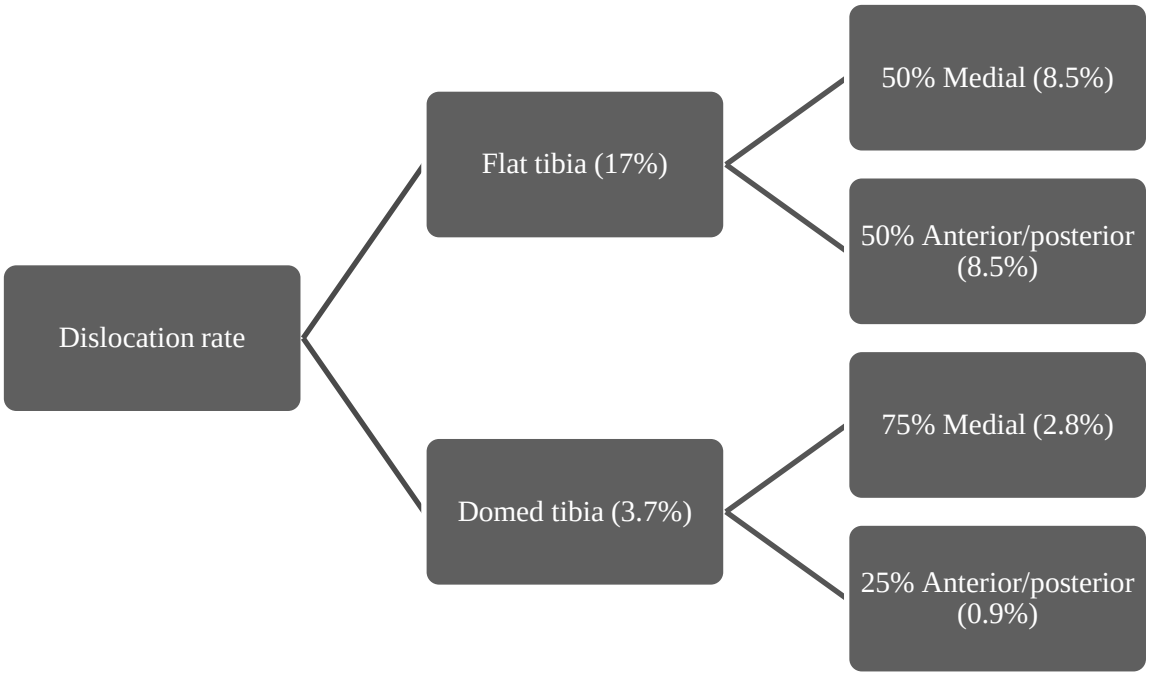


Figure 5.1. Flow chart showing clinical dislocation rates for dislocation directions, assuming dislocations with unspecified directions were antero-posterior.

Of the 17% bearing dislocation rate, medial dislocations and anterior/posterior dislocations each accounted for approximately 50% of the bearings that dislocated using the Oxford flat lateral implant. This equates to an equally distributed 8.5% medial dislocation rate and an 8.5% anterior/posterior dislocation rate. For the Oxford Domed Lateral (ODL) implant, of the 3.7% bearing dislocation rate, medial dislocations accounted for 75% of the bearings that dislocated, and the remaining 25% of the bearings that dislocated were anterior/posterior dislocations. This equates to a 2.7% medial dislocation rate and a 0.9% anterior/posterior dislocation rate.

5.4.2 Dislocation risk calculated computationally

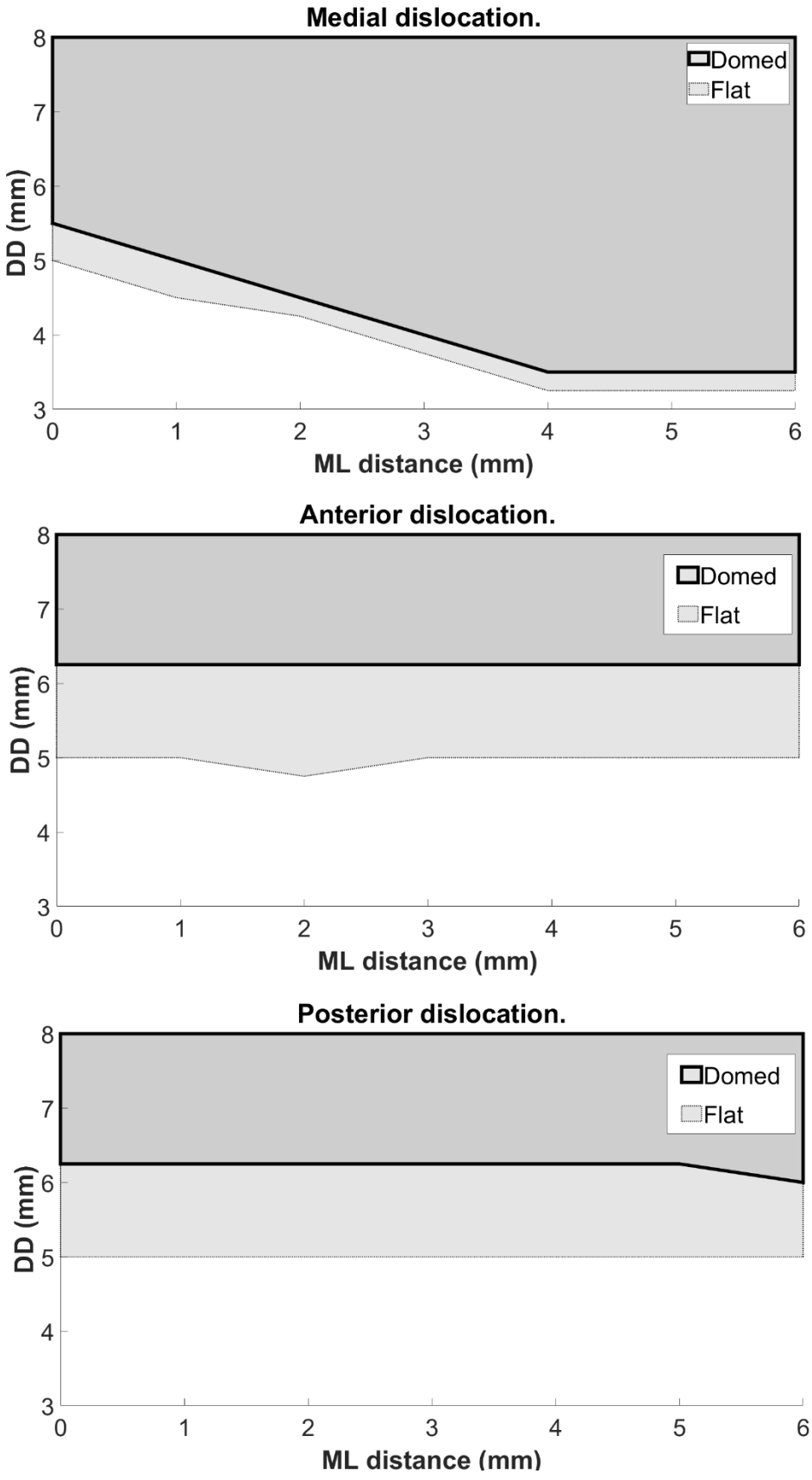


Figure 5.2. Medial, anterior or posterior dislocation: Oxford flat lateral versus ODL implant.

Using the dislocation analysis tool, medial, anterior and posterior dislocation risk was compared when using either the Oxford flat lateral versus ODL implant (Figure 5.2). For the flat tibial surface, for ML 0 – 6 mm, the DD medially was 5 – 3.25 mm and 5 mm anteriorly/posteriorly. For the domed tibial surface, for ML 0 – 6 mm, the DD medially was 5.5 – 3.5 mm and 6.25 mm anteriorly/posteriorly.

Changing the design of the tibial surface from a flat to a domed design increased the medial DD by 0.5 mm between 0 mm and 1 mm ML and 0.25 mm from 2 mm to 6 mm ML. Changing the design of the tibial surface from a flat to a domed design increased the anterior and posterior DD by 1.25 mm.

5.5 Discussion

5.5.1 Dislocation rates by dislocation direction

The systematic review results demonstrate that clinically, medial dislocation is the most common mode of dislocation for both the Oxford flat lateral implant and the ODL implant, accounting for 50% of dislocations that occurred using the Oxford flat lateral implant and 75% of dislocations that occurred when using the ODL implant. The inferred clinical dislocation rates for different dislocation direction results suggest that changing the design of the tibial component from a flat to a domed design reduced the clinical anterior/posterior dislocation rate from 8.5% to 0.9%, whereas the medial dislocation rate was reduced from 8.5% to 2.8%. It is noted that the Gunther et al. results was an update of the series originally reported by Goodfellow et al. in 1988 [122] on the Oxford phase I and II implants, where it appears that one additional dislocation had occurred between 1985 and 1991. In the Gunther et al. study, a total of six dislocations in fifty-three lateral unicompartmental replacements were reported [19] however, the direction of dislocation for each dislocation was not reported. Therefore, for the purpose of this study, it was decided to report the original Goodfellow et al. series [122] as it contained the information on dislocation directions. In the data collected, only some dislocations were clearly marked as anterior [6], while many reports did not mention explicitly whether dislocations occurred anteriorly or posteriorly [6, 7, 122]. It was found that in most published

articles in which the direction of dislocation has been reported, medial dislocations were explicitly stated, probably as it is the most common dislocation direction, so it was therefore assumed that the remaining dislocations occurred either anteriorly/posteriorly. For example, in the Kennedy et al. publication [6], there were two dislocations in which the mode was unknown, and so these could have been either medial, anterior or posterior dislocations. It was decided to add these to the anterior/posterior dislocations, since lateral dislocation do not occur clinically and had the dislocations been medial, they were likely to be recorded as medial dislocations, so caution is advised when reviewing these results. For the remaining studies, all the dislocation directions were explicitly stated.

5.5.2 Dislocation risk calculated computationally

Using the computational dislocation analysis tool, the minimum DD for each mediolateral distance was identified. It was found that using the Oxford flat lateral implant, for ML distances between 0 – 6 mm, the DD medially was between 5 – 3.25 mm and the DD anteriorly/posteriorly was 5 mm, whereas using the ODL implant, the DD medially was 5.5 – 3.5 mm and the DD anteriorly/posteriorly was 6.25 mm. Therefore, changing the design of the tibial component from a flat to a domed tibial surface increased the DD medially, anteriorly and posteriorly, however, the increase in DD was larger for anterior/posterior dislocation than for medial dislocation. Based on the principle that the larger the DD, the lower the risk of bearing dislocation, the results suggest that changing the design of the tibial component from a flat to a domed tibial surface decreased the anterior/posterior dislocation risk more than the medial dislocation risk. However, as the actual amount of distraction that occurs in the lateral compartment of the knee depends on many factors such as the position of the knee, the laxity of the ligaments [164] and the loading on the knee [32], the clinical dislocation rates may vary from these results.

5.5.2.1 *Dislocation risk calculated computationally: comparison with Weston-Simons et al.*

Interestingly, when the results from the computational dislocation analysis tool assessing flat versus domed tibial components were compared with the results of flat versus domed tibial components previously reported by Weston-Simons et al. [10] (Table 5.3), the results were again quite different. This study found that at 1 mm ML, using the Oxford flat lateral implant, the DD medially was 4.5 mm and the DD anteriorly/posteriorly was 5 mm, whereas using the ODL implant, the DD medially was 5 mm and the DD anteriorly/posteriorly was 6.25 mm. In contrast, Weston-Simons et al. [10] reported that for a 3 mm bearing positioned 1 mm ML, using the Oxford flat lateral implant, the DD medially was approximately 3.5 mm and the DD anteriorly/posteriorly was approximately 4 mm, whereas using the ODL implant, the DD medially was 5.9 mm and the DD anteriorly/posteriorly was 8.1 mm.

Table 5.3. Comparing the current results with Weston-Simons et al. [10]: Distraction to Dislocation (DD) medially or anteriorly/posteriorly using either the Oxford flat lateral implant or the Oxford Domed Lateral (ODL) implant measured when the bearing is 1 mm from the tibial wall.

	Oxford Implant used for testing	Distraction to Dislocation (DD) (mm)	
		Medially	Anteriorly/posteriorly
This study	Oxford flat lateral implant	4.5	5
	Oxford Domed Lateral (ODL) implant	5	6.25
		Medially	Anteriorly
Weston-Simons et al. [10]	Oxford flat lateral implant	3.5*	4*
	Oxford Domed Lateral (ODL) implant	5.9	8.1

*Results are approximate readings taken from the results reported by Weston-Simons et al. [10].

Therefore, using the Oxford flat lateral implant, the DD results from this study are 1 mm higher for both medial and anterior/posterior dislocation than those reported by Weston-Simons et al. [10]. When using the ODL implant, the DD results from this study were 0.9 mm lower for medial dislocation and 1.85 mm lower for anterior/posterior dislocation than those reported by Weston-Simons et al. [10]. This suggests that compared with dislocations tested using the computational dislocation analysis tool, dislocations using the mechanical rig in Weston-Simons et al.'s publication

[10] were easier when using the Oxford flat lateral implant but more difficult when using the ODL implant. These differences are inconsistent, and do not make much sense. It is not clear from Weston-Simons et al.'s publication [10] what definition was used for "dislocation" using the Oxford flat lateral implant, nor is it clear how the dislocations were achieved (e.g. it is not clear what pathway was used to dislocate the bearing or what constituted a dislocation). Given the analysis in Chapter 3, the inconsistent differences between the results are perhaps expected since it is clear that DD values measured using the mechanical rig were only as reliable as the definitions used for dislocation and the sequence of movements to achieve dislocation that were defined.

5.5.3 Dislocation rate versus dislocation risk

A closer analysis of the geometry of the implant may explain why the results in this study show minimal difference to the medial DD values between the flat tibial design and the domed tibial design. Changing the tibial surface from a flat to a domed shape increased the entrapment of the mobile bearing, however, at the medial side of the bearing (Figure 5.3), it can be seen that the amount of bearing entrapment increase is larger at the anterior/posterior extremities (distance between line 1 and line 3), than at the centre of the bearing (distance between line 1 and line 2). Measuring these distances computationally, the distance between line 1 and line 2 was 0.9 mm whereas the distance between line 1 and line 3 was 2.6 mm.

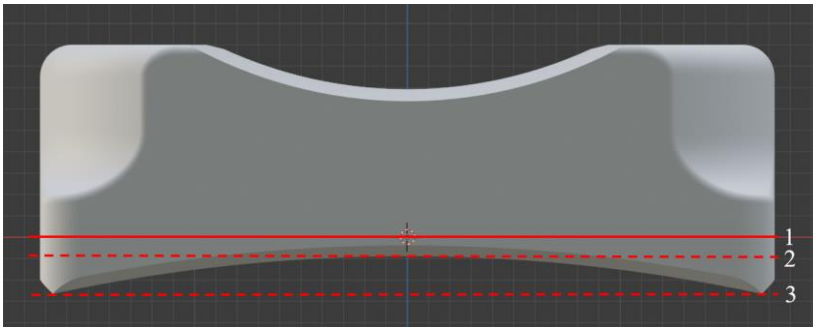


Figure 5.3. The effect of changing the tibial component from a flat to a domed tibial surface. Line 1 is located in the centre of the bearing at the peak of the dome, line 2 and line 3 are located on the medial side of the bearing, where line 2 aligns with the vertical centre and line 3 aligns with the maximum heights located at the anterior/posterior extremities.

While it may be expected that a maximal increase of 2.6 mm would result in an increase in DD medially of more than 0.25 mm, it is also noted that when the design of the tibial surface was changed from a flat to a domed tibial surface, the shape of the tibial wall was also altered correspondingly with the underside of the bearing having the same profile as the tibial wall (Figure 5.4). While this was necessary to avoid metal-on-metal collision between the femoral and tibial components in flexion, this may have negated some of the efforts gained in changing the shape of the tibial surface by making the effective wall heights for the flat versus the domed implants very similar. Therefore, given that for a medial dislocation to occur the bearing just needs to sublux above the tibial component wall, this possibly explains why medially the increase in bearing entrapment only resulted in a 0.25 mm increase in DD medially.

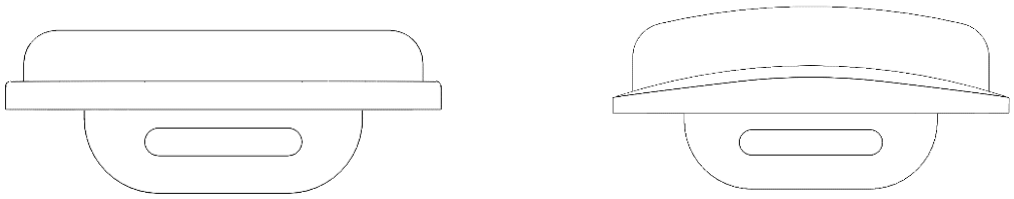


Figure 5.4. Flat (left) versus domed (right) tibial component wall.

Taken together, the results suggest that as initially intended, changing the design of the tibial component from a flat to a domed tibial surface successfully reduced the overall dislocation risk. Further, previous studies have also reported that the ODL implant improves the kinematics, flexion and function [114], suggesting that the ODL has further advantages other than to reduce the dislocation risk and similar results were reflected in Chapter 2, Figure 2.7. However, regarding the biggest issue with the ODL implant which is bearing dislocation, while the proportion of medial dislocations increased, the overall dislocation risk decreased with the reduction in the anterior/posterior dislocation risk being greater than the reduction in the medial dislocation risk. The theoretical risk of dislocation measured computationally using the dislocation analysis tool provides a plausible explanation for the results seen clinically: introducing a domed tibial surface increased the anterior/posterior DD more than the medial DD, which in turn increased the anterior/posterior DD more than the medial DD (0.25 mm versus 1.25 mm) and may be the reason why the

anterior/posterior dislocation rate decreased from 8.5% to 0.9%, while the medial dislocation rate decreased from 8.5% to 2.8%. This also explains why proportionally, medial dislocations appear to account for more dislocations using the ODL implant rather than the Oxford flat lateral (50% vs 75%).

5.5.4 Study limitations

The main limitation in this chapter was the low number of studies included for analysis of clinical dislocation rate. For the Oxford flat lateral implant, only two studies were included with a total of 92 knees and for the ODL implant, only three studies were included with a total of 391 knees. Further, it was clear that it is currently not routine practice to report on the dislocation directions [7, 19], so when information was available, it was often incomplete. Another limitation in this study is that the computational results for the flat lateral implant were compared to the flat lateral results reported by Weston-Simons et al. [10]; further analysis could directly compare the flat lateral computational results to direct testing using the mechanical rig.

5.6 Conclusion

Changing the design of the tibial component from a flat to a domed tibial plateau increased the DD anteriorly/posteriorly by 1.25 mm compared with just 0.25 mm medially because introducing a domed tibial surface increased the anterior/posterior DD more than the medial DD. This may explain why clinically, it appears that changing the design of the tibial component from a flat to a domed design had a greater effect at reducing the anterior/posterior dislocation rate (8.5% to 0.9%) than on the medial dislocation rate (8.5% to 2.8%). Future work should aim to focus on efforts to reduce the risk of medial dislocation, which is the main issue with the ODL implant.

Chapter 6 DESIGN MODIFICATIONS TO THE ODL IMPLANT TO REDUCE MEDIAL BEARING DISLOCATIONS

6.1 Introduction

It is clear from this thesis that changing the design of the ODL implant can influence the dislocation rate. As demonstrated in Chapter 2, changing the design of the Oxford knee from a flat tibial plateau to a domed tibial surface successfully reduced bearing dislocation risk from 17% to 3.7% [9]. However, despite this change, dislocation remains an issue clinically [9]. Therefore, in an attempt to reduce the dislocation risk, this chapter presents a range of design modification ideas proposed with the goal to reduce the bearing dislocation risk without compromising on the functionality and performance of the existing implant. The chapter presents all the design ideas and how these were generated. The chapter also explains the rationale behind the decisions to arrive at the final design modifications selected for in silico testing.

6.2 Aims

In this chapter, the aim was to generate new ideas on how to re-design the ODL implant components to retain the functionality and overall performance of the ODL implant, while reducing the risk of mobile bearing dislocation. Another aim was to establish a systematic way to filter these design ideas in order to identify the feasible design modifications, which would then be selected for further testing.

6.3 Methods

There were two methods identified to improve the design of the implant components to reduce the likelihood of dislocation of the mobile bearing. The first method focused on reducing the minimum distance between the femoral component and the tibial wall. The second method focused on re-designing the bearing to resist medial dislocation e.g. increasing the bearing entrapment to resists dislocations medially/laterally. “Other ideas”, which are more radical design changes are also presented in this chapter.

Any of the three components (femoral component, tibial component or the mobile bearing) that comprise the overall ODL implant could be re-designed. For each component, the mechanical design and geometry was analysed to identify specific design features that could be altered to reduce the likelihood of mobile bearing dislocation. In the first instance, new design ideas were brainstormed without any constraints. Design ideas were subsequently filtered based on the user needs and implant safety – each design was assessed for whether it satisfied the user need or not (Table 6.3). If the user need was satisfied, a tick was indicated. Since it was decided that all new design ideas can not affect the biomechanics of the implant, any design with did not fulfil this user need was immediately eliminated. The final tally assessed the total number of user needs that were satisfied. Designs that did not affect the biomechanics and which had a remaining score higher than 6 was selected for further testing. Chosen concepts were designed initially using hand sketches on paper and actualised using SOLIDWORKS 2020 (Dassault Systemes SOLIDWORKS Corp., Waltham, Massachusetts, USA) and concurrently prototyped to obtain physical models. Additionally, clinical input was imperative to selecting the final design.

6.4 Results

Together with orthopaedic surgeon, Professor David Murray, and engineer, Associate Professor Stephen Mellon, Table 6.1 presents the user needs to facilitate selection of final designs. The features that could be altered and the alterations that are likely to affect the mobile bearing dislocation risk were identified and are presented in Table 6.2. Following this, the proposed design changes are presented in Table 6.3. Finally, the selected design modifications are segregated by component: tibial (Table 6.4), femoral (Table 6.5) and mobile bearing (Table 6.6).

Table 6.1. User needs.

Original user needs [90]
Device is intended for the replacement of diseased or worn articular surfaces.
Device must retain as much of the native ligaments and muscles in the knee joint as possible.
All retained soft tissues must be restored to natural tension.
Components must be shaped to enable distraction, sliding and rolling motion between the bones of the knee joint.
Components should apply only compressive stresses to the juxtra-articular bone.
The materials used and contact area between component surfaces should be sufficiently strong to withstand the pressure when the knee is loaded and in compression.
New user needs
The new design must reduce the dislocation risk when compared with the existing ODL implant design.
The new design should meet or exceed the performance and functionality of the current ODL implant design, including: - Treat end-stage lateral unicompartmental knee osteoarthritis - Be inserted minimally invasively - Retain the native knee ligaments - Restore normal knee kinematics - Enable posterior roll back of the femoral component relative to the tibial component - Reduce bearing wear when compared to fixed bearing designs - Have sufficient bearing entrapment to reduce the problem of bearing dislocation
The new femoral/tibial components should still fit the distal femur and proximal tibia, respectively. The bearing must fit between the femoral and tibial components.
The bearing wear rate should not be worse than the current wear rate.
The two metal components are not to contact or collide at any point as this may result in metallosis.
The implant should retain current biocompatibility and sterilizability.
The design of the implant should be manufacturable.
The surgical technique should not alter considerably from the current surgical technique however minor alterations to the existing surgical technique or instrumentation are possible.

Table 6.2. ODL implant components: features to alter/target for design changes to reduce mobile bearing dislocation.

Component	Features that cannot be altered	Features that can be altered but have no effect on bearing dislocation	Features to target

Femoral	• The spherical shape of the femoral component.
	• The spherical design of the femoral component.
		• Specific features of the central posts that anchor the implant into the femur.
		  ○ The number of femoral posts
		  ○ The material of the post
		  ○ The surface profile of the non-contacting surface of the femoral component.
			• The surfaces which contact the bearing:
			  ○ The radius of curvature of the femoral component.
			  ○ The width of the femoral component.

Tibial	• The overall D-shape.
	• The domed feature of the tibial plateau.
	• The presence of a tibial wall.
		• Any features located on the underside of the tibial component (e.g. the keel or the cut on the underside of the component).	• The surfaces which contact the bearing:
			  ○ The radius of curvature of the domed tibial plateau.
			  ○ The height and shape of the wall.

Mobile bearing	• The biconcave design of the mobile bearing.	• The metal radiographic rod holes. • The circumferential wall edges of the bearing. • The step design of the medial anterior and posterior corners of the bearing.	• The radius of curvature of the inferior and superior surfaces of the bearing. • The width and height of the bearing.

Table 6.3. Design ideas to reduce dislocation of the mobile bearing.

Method	Idea	Affects <1 component?	Easily manufactured?	Retains existing mechanical axis?	Retains bearing wear rate?	Retains metallosis chance?	Does not require new material?	Retains surgical technique?	Retains existing instrumentation?	Retains biomechanics?	Overall total	Selected for further testing?
Reduce the min distance between the metal components	Increase the height of the wall without increasing metallosis chance. • Nominally • Central portion only	✓	✓	✓	✓	✓	✓	✗	✓	✓	**8**	**Y**
	Add a lip to the top of the wall which can be angled medially/laterally (coronal plane).	✓	✗	✓	✓	✗	✗	✓	✗	✗	**4**	**N**
	Increase the thickness of the tibial wall in the inner side that contacts the mobile bearing	✓	✓	✗	✓	✓	✓	✓	✗	✗	**6**	**N**
	Level out the tibial wall curvature in sagittal plane to	✓	✓	✓	✓	✗	✓	✓	✓	✗	**7**	**N**

	increase the capture of the winged edges on the anterior medial or posterior medial corners of the mobile bearing. • Uniformly • Anterior only • Posterior only										
	Flatter curvature to level out tibial wall but not fully horizontal. • 90 degrees • 100 degrees • 120 degrees	✓	✓	✓	✓	✗	✓	✓	✓	✗	**7**	**N**
	Angle the tibial wall inwards towards the femoral component.	✗	✗	✓	✓	✗	✓	✓	✗	✓	**5**	**N**
	Increase the width of the femoral component (spherical extension). • Medially only	✓	✓	✓	✓	✓	✓	✓	✓	✓	**9**	**Y**

	• Laterally only • Uniformly											
Design the bearing to resist medial dislocation	Shift the centre of curvature of the inferior dome on the mobile bearing to increase curvature depth. • Medially only • Laterally only	✗	✗	✓	✓	✓	✓	✓	✗	✓	**6**	**Y**
	Increase the width of the inferior surface of the mobile bearing to increase the wing feature. • Uniformly • Medially only • Laterally only	✓	✓	✓	✗	✓	✓	✓	✗	✗	**6**	**N**
	Increase the width of the mobile bearing to increase the bearing entrapment. • Laterally only	✓	✓	✓	✗	✓	✓	✓	✓	✓	**8**	**Y**

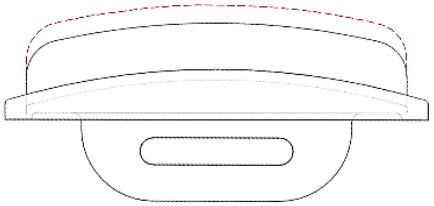
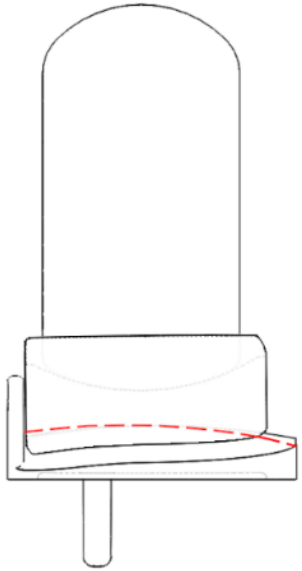
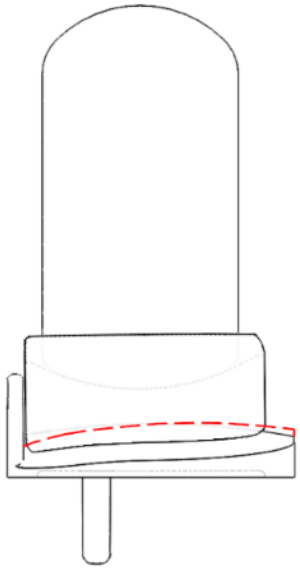
	• Medially only • Uniformly											
Other ideas	Merge the femoral component and bearing into one.	✗	✗	✓	✗	✓	✓	✗	✗	✗	**3**	**N**
	Add a medial wall onto the bearing.	✓	✗	✓	✓	✓	✓	✓	✗	✗	**6**	**N**
	Add a connection (string or some sort of anchor) between the mobile bearing and the femoral component: • Centrally • Medially • Anteriorly • Posteriorly • Laterally	✗	✗	✓	✓	✓	✗	✗	✗	✗	**3**	**N**
	Add a connection between the tibial wall to the closest point of the femoral component.	✗	✗	✓	✓	✓	✗	✗	✗	✗	**3**	**N**

	Add a connection to constrain the mobile bearing to the tibial component. String/suture-like material added on: • Underside of the bearing to the the tibial plateau. • Either corner of the bearing to the tibial component. • Centrally on the tibial wall to the medial edge of the mobile bearing.	✗	✗	✓	✓	✓	✗	✗	✗	✗	**3**	**N**
	Add a connection to secure the bearing to the nearby bone structures e.g. via suturing	✗	✗	✓	✗	✓	✗	✗	✗	✗	**2**	**N**
	Increase the radius of curvature of the	✗	✗	✓	✓	✓	✓	✓	✗	✗	**5**	**N**

	femoral component to increase bearing entrapment. • Coronal plane only → "toroid dome" - retains functionality in the sagittal plane • Overall											
	Increase the radius of curvature of the domed tibial plateau to increase bearing entrapment. • Coronal plane only → "toroid dome" - retains functionality in the sagittal plane • Overall	✗	✗	✓	✓	✓	✓	✓	✗	✓	**6**	**Y**

	Add a lateral lug to the bearing to increase the DD medially • Interacting with the tibial component? • Interact with the femoral component?	✗	✗	✗	✓	✓	✓	✗	✗	✗	**3**	**N**
	No tibial wall at all? So that there is nothing to get stuck on?	✓	✓	✓	✓	✓	✓	✓	✓	✗	**8**	**N**
	A completely plastic tibial wall?	✓	✗	✓	✗	✓	✗	✗	✗	✗	**3**	**N**
	A deformable bearing	✓	✗	✓	✓	✗	✗	✗	✗	✗	**3**	**N**
	Weighted bearing	✓	✗	✓	✗	✗	✗	✓	✓	✗	**4**	**N**
	Magnetic bearing	✓	✗	✓	✗	✗	✗	✓	✓	✗	**4**	**N**

Table 6.4. Final tibial component modifications selected for testing. NOTE: components depicted with the knee in flexion.

Design modification number	**1**	**2**	**3**
Design modification	Increase the tibial wall height (1 mm, 2 mm, 3 mm or 4 mm).	Shift the centre of curvature of the dome medially	Shift the centre of curvature of the dome laterally
Schematic representation			
Other notes	Corners may have a smaller radius. For each millimetre of wall height increase, the bearing has to be 1 mm thicker†.	N/A	N/A

†The thinnest bearing available is nominally 3 mm thick.

Table 6.5. Final femoral component modifications selected for testing. NOTE: components depicted with the knee in flexion.

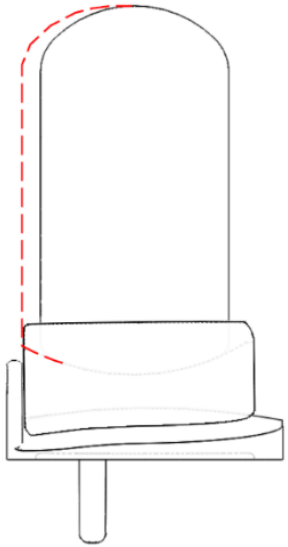
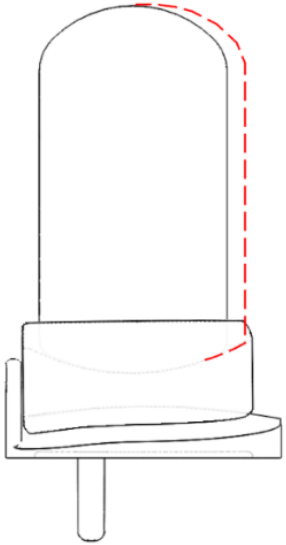
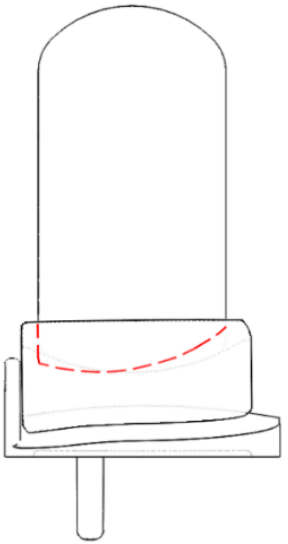
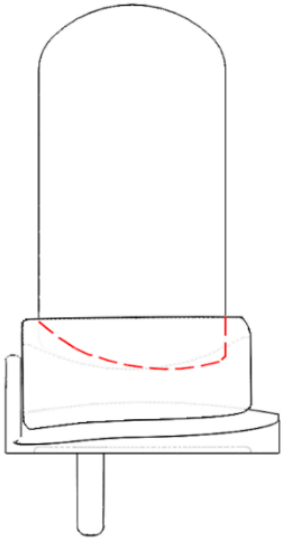
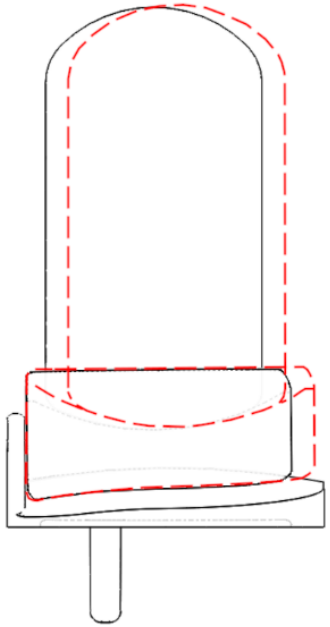
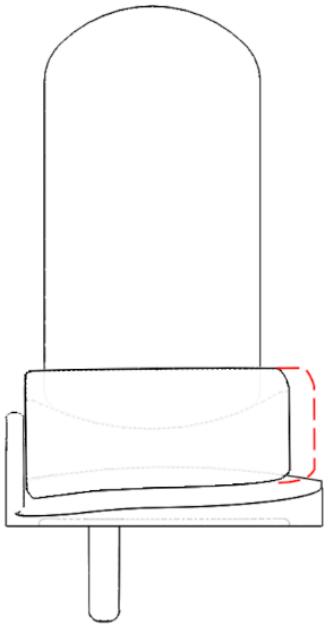
Design modification number	**1**	**2**	**3**	**4**
Design modification	Wider medially (2 mm)	Wider laterally (2 mm)	Shift the centre of the sphere medially (3 mm)	Shift the centre of the sphere laterally (3 mm)
Schematic representation				
Other notes	N/A	N/A	Tibial wall height must be reduced.	Tibial wall height can be increased.

Table 6.6. Final mobile bearing component modifications selected for testing. NOTE: components depicted with the knee in flexion.

Design modification number	**1**	**2**
Design modification	Wider bearing medially (1.6 mm)	Wider bearing laterally (2 mm)
Schematic representation		
Other notes	Bearing changes affect femoral start position.	N/A

6.5 Discussion

In this chapter, twenty-three new implant modification ideas were presented. Each design modification would affect either the femoral component, tibial component, the mobile bearing or a combination of these components. While the exact cause and mechanism for medial bearing dislocation in the ODL implant is not well understood [9], this thesis has demonstrated that bearing dislocation risk is related to the relative position of the femoral and tibial components. Taking into account the user needs, design modifications which alter the relative geometries between the components to optimise the bearing entrapment or to reduce the minimum gap between the femoral and tibial components were conceived and selected for further testing. It is believed that these changes would reduce the bearing dislocation risk.

Bearing dislocation was treated as a static geometric problem and focused on modifying the design of the ODL implant to reduce the dislocation rate. However, given that the knee is complex in structure, highly mobile and kinematically complex, bearing dislocation is also likely affected by many other factors inside the knee joint which may contribute to dislocation of the mobile bearing. However, previous attempts to address the problem of bearing dislocation through modifications to the geometry of the device have successfully reduced the dislocation risk: changing the design of the tibial component from a flat tibial surface to a domed tibial surface increased the bearing entrapment [159], reducing the bearing dislocation rate by approximately 80% [9]. Furthermore, as the problem of bearing dislocation is a geometric problem with the bearing escaping from between the femoral and tibial components, it seems intuitive that increasing the bearing entrapment, or reducing the gap through which the bearing is escaping medially should reduce the risk of bearing dislocation. Finally, while it has been assumed that bearing dislocation is a static problem, bench testing and in silico testing will test a wide range of physiologically relevant positions. Subsequently, cadaveric and clinical testing are required in order to ensure that the other factors that may affect dislocation are also checked, and that the new design meets the user needs as set out in the user needs.

Changes to the design of a multi-component implant rarely only address the issue in isolation, however, the key is to identify a set of design changes that does not alter or improves on the performance of the existing design. As an example, the primary impetus for the design and introduction of the ODL implant was to increase the bearing entrapment thereby reducing the risk of bearing dislocation. The ODL implant design also improved the anatomic relevance of the implant to the lateral knee by mimicking the native lateral knee convex tibial shape, and the normal knee kinematics by allowing greater rollback of the femoral component relative to the tibial component [159].

The integrity of any implant idea or orthopaedic innovation is based on the strength of the team behind the innovation. Many informal discussions were held with clinicians and researchers in the orthopaedic community in Oxford, Cambridge, the wider UK and international bodies throughout the duration of the project to refine the clinical need. Further, Professor Murray and Associate Professor Mellon are both experts in the fields of orthopaedic surgery and orthopaedic engineering, respectively, with extensive innovation experience particularly in relation to the design of the Oxford Knee.

6.6 Conclusions

In this chapter, twenty-three new implant modification ideas have been proposed for the ODL implant in order to reduce the risk of medial bearing dislocation. The user needs were also presented along with the decision making table to demonstrate how these were taken into account when selecting the final potential design modifications. Out of the twenty-three proposed ideas, nine designs modifications were selected for further testing (three tibial, four femoral and two bearing modifications).

Chapter 7 THE NEW ODL IMPLANT DESIGN TO RESIST MEDIAL DISLOCATIONS – A PROPOSAL

The results of the higher wall implant from this chapter have been published in a peer reviewed journal[6]

7.1 Introduction

In previous chapters, it has been shown that distraction in the lateral knee increases the risk of bearing dislocation. Further, as shown in both Chapter 2 and in Chapter 5, the design of the implant can be altered to reduce the dislocation rate. In this chapter, the aim was to introduce changes that can be made to the ODL implant so that even when the knee distracts, the risk of bearing dislocation is reduced. Even though anterior and posterior dislocations are possible they rarely occur clinically so the risk of AP dislocation is considered acceptable. Therefore, if the design of the implant could be modified so that the medial dislocation matches the anterior/posterior dislocation risk, this would probably reduce the clinical dislocation rate to an acceptable level.

7.2 Aims

The aim of this chapter was to test the final design modifications identified in Chapter 6 to determine the optimal implant design modifications that would successfully reduce the risk of bearing dislocation.

[6] I. Yang, Jonathan D. Gammell, D. Murray, S. Mellon (2021), The Oxford Domed Lateral Unicompartmental Knee Replacement implant: Increasing wall height reduces the risk of bearing dislocation. Proceedings of the Institution of Mechanical Engineers, Part H: Journal of Engineering in Medicine. doi:10.1177/09544119211048558

7.3 Methods

7.3.1 Designing and modelling the modified components

Modified CAD 3D models of ODL components depicted in Table 6.4, Table 6.5 and Table 6.6 were modelled were made using SOLIDWORKS 2021 (Dassault Systemes SOLIDWORKS Corp., Waltham, MA, USA). More specifically, for higher wall testing, the components modelled were a medium size femoral component, size C dome lateral tibial component with normal wall height and wall heights increased by 1 mm, 2 mm, 3 mm or 4 mm, and, a medium size domed lateral bearings with nominal thickness of 3 mm, 4 mm, 5 mm, 6 mm and 7 mm.

In addition to assessing the effect of increasing wall height, the effect of individual design modifications to each component on the risk of medial mobile bearing dislocation was also assessed. To test the effect of shifting either the femoral or tibial dome centres, or the effect of increasing the width of the femoral or bearing components, the components modelled were: tibial components with the dome centre shifted 3 mm (medially or laterally), femoral components with 2 mm width increase (medially or laterally), femoral components with the femoral sphere centre shifted 3 mm (medially or laterally) and bearing components with the width increased either 1.6 mm medially, or 2 mm laterally. Finally, to test the new design, which featured a combination of the selected design modifications, modified femoral, tibial and bearing components were also modelled in SOLIDWORKS. From the SOLIDWORKS part models, STL files were created. The STL models were inserted into the robotics dislocation analysis tool to assess the dislocation risk.

For each component, individual modifications were made without changing the design of the remaining two components. However, some design modifications could not be done in isolation and required changes to other components as well. For example, shifting the centre of origin of the sphere 3 mm either medially or laterally also required the tibial component wall height to be changed accordingly to ensure a minimum distance of 2 mm between the metallic femoral and tibial components. This would prevent the metal components from hitting each other which, if it does occur, may result in clinical complications relating to metal debris. In the long term, there will be

wear of the polyethylene bearing, meaning that when there is rotation within the knee, the femoral component may be above the wall of the tibial component, and if the bearing is worn, metal-on-metal collision is more likely. Therefore, shifting the femoral sphere centre medially required the tibial wall height to be reduced. While not compulsory but beneficial to preventing dislocations, shifting the femoral sphere centre laterally meant that the tibial wall height could be increased until it was 2 mm away from the femoral sphere. To determine the amount of height increase or decrease of the tibial component wall, the distance between the metallic femoral component and the top of the tibial wall was measured. To do this, a sphere of equivalent radius to the femoral component was placed where the femoral component would sit if the bearing were flush against the tibial wall. The closest 3D distance between the tibial wall and the femoral sphere was measured using the MATLAB Euclidean distance measurement [187].

7.3.2 Testing the modified components using the dislocation analysis tool

For higher wall testing, the effect of each design modification on the risk of medial and AP dislocations were assessed using the dislocation analysis tool. For the remaining modifications, medial dislocations were assessed. Following results, the design modifications that successfully reduced the bearing dislocation risk were identified, and combined into the new design. Using the robotics dislocation analysis tool the DD results for the new implant design were compared with the original implant design.

Given the congruent nature of the components, the position of the femoral component relative to the tibial component controlled the position of the bearing. In this chapter, the configuration refers to the specific combination of components used for testing as well as the defined mediolateral (ML) distance. For each configuration, the starting position was setup with the femur in the position where the bearing would be flush against the tibial component wall with the unaltered components. The tibial component was then moved medially in 0.25 mm increments so the ML distance between bearing and wall increased from 0.00 mm to 6.00 mm (see Figure 7.1).

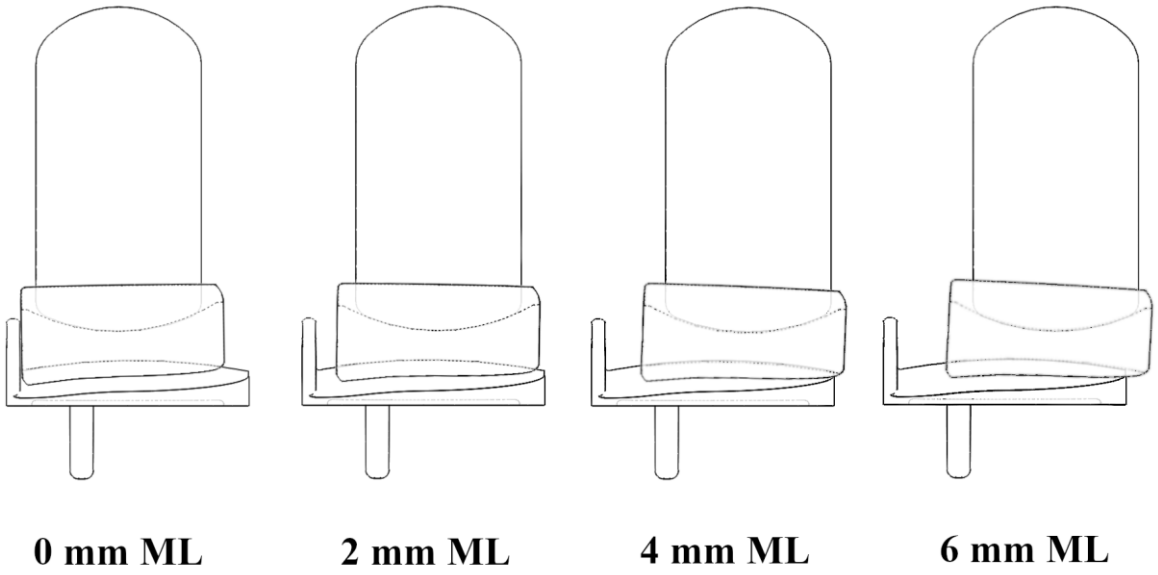


Figure 7.1. Series of images depicting 0 mm, 2 mm, 4 mm and 6 mm ML.

At each position, the femoral component was then moved vertically away from the tibial component (0.00 – 8.00 mm) in 0.25 mm increments. The dislocation analysis tool used the Rapidly-Exploring Random Trees (RRT) algorithm [182] to automate the motion of the mobile bearing by randomly sampling incremental steps (0.25 mm) in the space available between the femoral and tibial components such that the bearing may move from the starting non-dislocated position, until a dislocated position has been found. For each search, the algorithm searched each ML translation distance starting from the highest vertical distraction value (8 mm). For each configuration, the RRT was allowed to run for 270 seconds and each search was repeated 10 times. To ensure that a non-dislocator was in fact not dislocating as a result of the geometric design of the components and its relative position, if a non-dislocator was identified, the search was repeated but the search time was extended to 405 seconds and the search was repeated 25 times. A bounding box was used to confine the search region. A medial dislocation occurred when the centre of mass of the bearing entered the "goal area", defined by the anterior and posterior edges of the tibial component, medially by the tibial wall and laterally by the halfway distance between the midpoint of the tibial component and the tibial wall. An anterior/posterior dislocation occurred when the bearing entered the "goal area", defined medially by the tibial wall, laterally by the lateral edge of the tibial component, and the region 10

mm beyond the anterior/posterior edges for anterior/posterior dislocation, respectively. When a dislocation was detected, the DD was recorded i.e. the amount the femoral component had been moved vertically away from the tibial component.

7.3.3 Validating the computational results using the mechanical rig

Following a similar procedure outlined in *Section 3.3.2.1 Modifying the rig to mimic knee flexion* in Chapter 3, the modified femoral, tibial and mobile bearing components were CAD designed in SOLIDWORKS and 3D printed in Polylactic Acid (PLA). With the help of James Fisk at the Institute of Biomedical Engineering (IBME) at the Department of Engineering Science at the University of Oxford, the femoral component was modified to fit in the rig (Figure 7.2).

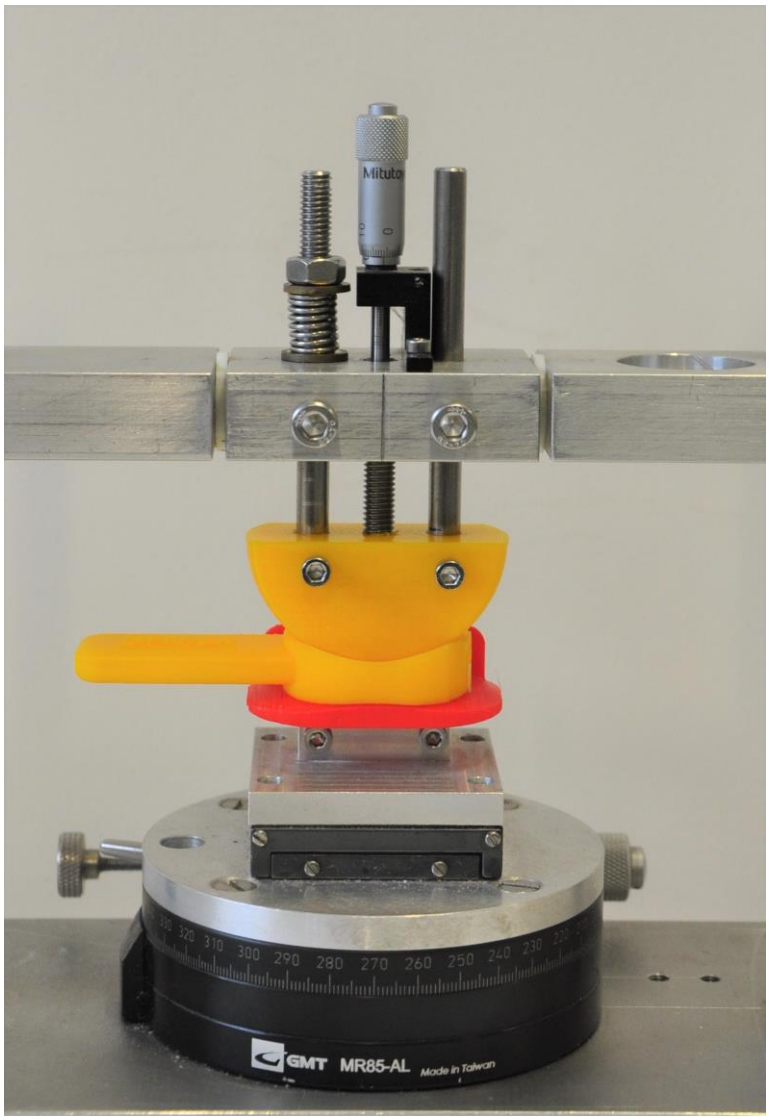


Figure 7.2. The new femoral, tibial and mobile bearing components in the mechanical rig.

The testing parameters used are shown in Table 7.1. Testing assessed a total of 259 implant positions for dislocation. The starting position was chosen with the bearing flush against the wall of the tibial component as per Figure 3.6 and central on the tibia with 0° rotation and 0 mm ML translation. To set up for testing, the configuration was determined by setting the bearing AP position, tibial rotation angle and mediolateral translation distance (the gap between the bearing and the tibial wall). For each configuration, the dislocation risk was assessed by recording the DD. The assessed vertical distraction ranged from 0.00 – 8.00 mm in 0.25 mm increments. The vertical distraction testing was conducted to an accuracy of 1/4 mm and mediolateral translation accuracy was chosen to an accuracy of 1 mm. All data were reported to two decimal places to allow comparison.

Table 7.1. Testing parameters.

Testing parameter	Testing range (increment)/values
Mediolateral translation (increment)	0 – 6 mm (1 mm increments)
Vertical distraction (increment)	0.00 – 9.00 mm (0.25 mm increments)

To assess medial dislocation, the mobile bearing was manually manipulated using the handle attached to the modified mobile bearing. To dislocate the bearings medially, the bearing was allowed to dislocate via any sequence of movements that would allow the bearing to escape medially. Using the mechanical rig, medial dislocations were recorded in a spreadsheet in Excel (version 2016, Microsoft, Redmond, Washington, U.S.).

7.4 Results

7.4.1 Higher tibial wall results

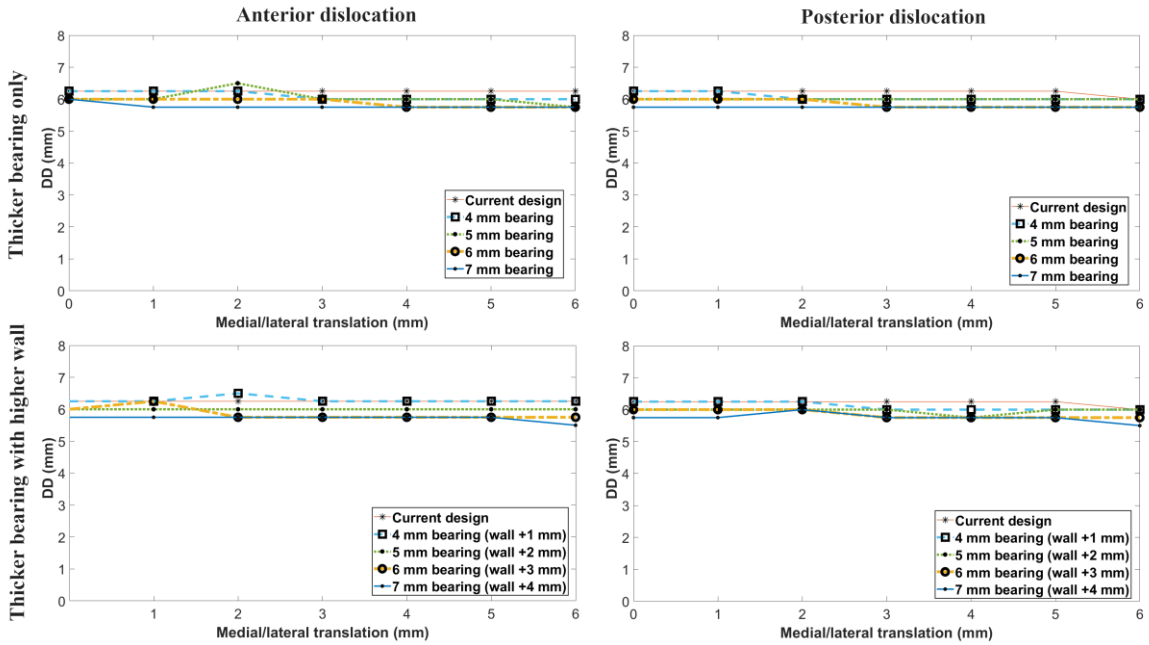


Figure 7.3. The effect of bearing thickness and increased tibial wall height on anterior and posterior dislocation. NOTE: Current design refers to the 3 mm bearing with standard tibial component wall height.

The DD for an anterior or posterior dislocation was identical. It progressively decreased with increasing bearing thickness: with a 3 mm bearing the DD was 6.25 mm and with a 7 mm bearing, the DD was 5.75 mm. The DD anteriorly/posteriorly was independent of wall height. Increasing the tibial component wall heights correspondingly with the bearing thicknesses, did not affect the DD anteriorly/posteriorly (Figure 7.3).

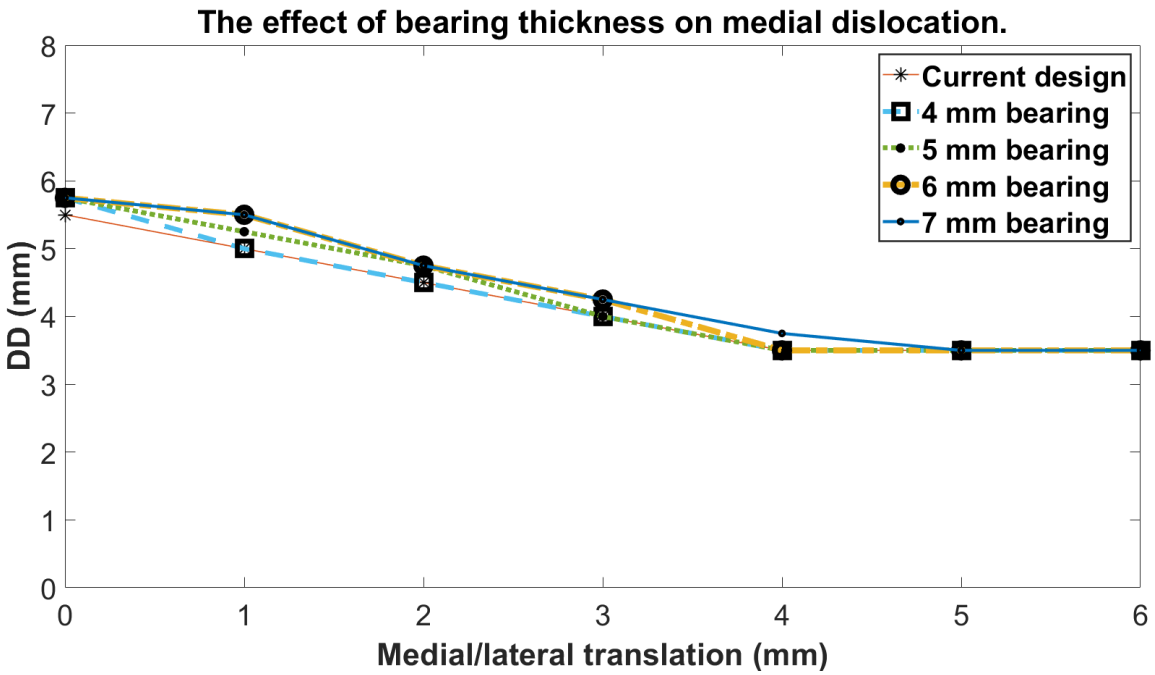


Figure 7.4. The effect of bearing thickness on medial dislocation. NOTE: Current design refers to the 3 mm bearing with standard tibial component wall height.

For a medial dislocation, with a 3 mm bearing and the standard wall height, the DD was 5.50 mm when the bearing was touching the wall (Figure 7.4). As the femur and bearing were moved laterally the DD progressively decreased becoming 4.50 mm with the bearing 2 mm from the wall and 3.50 mm with the bearing 4 mm from the wall. Further lateral movement of the femur and bearing (up to 6 mm) did not alter the DD.

For medial dislocation using thicker bearings increased the DD a small amount. For example with the bearing positioned flush against the wall, the DD increased by about 0.25 mm when the bearing thickness increased from 3 mm to 6 mm. When the bearing was more than 4 mm from the wall bearing thickness had no effect on the DD medially (Figure 7.4).

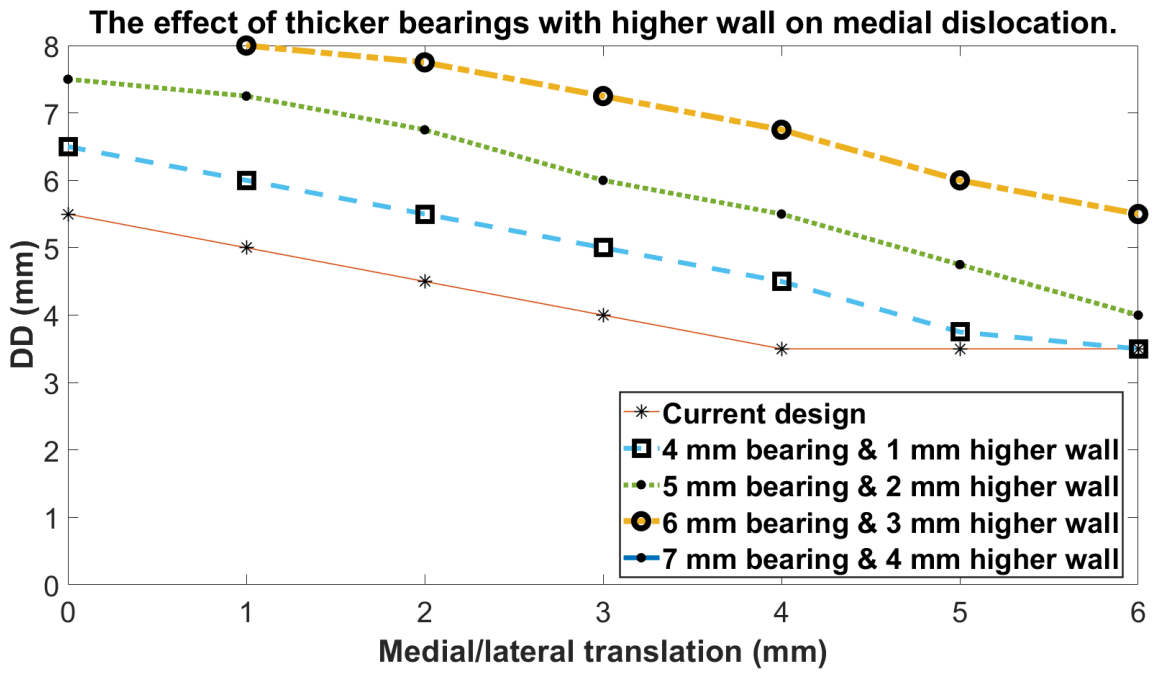


Figure 7.5. The effect of increasing wall height and using the corresponding thinnest bearing possible on medial dislocation. NOTE: "Current design" refers to the 3 mm bearing with standard tibial component wall height. There is no 7 mm line plotted because in the parameters tested, the 7 mm bearing did not dislocate.

For medial dislocation increasing the height of the wall increased the DD. For each 1 mm the wall height was increased, the DD increased by about 1 mm. This increase was independent of the ML position of the femur and bearing and the bearing thickness (Figure 7.5). The only exception to this was when the bearing was 5 mm or 6 mm from the tibial component wall, when the increase was smaller. If the wall height was increased by 2 mm then the DD with bearing flush with the wall, 2 mm from the wall and 4 mm from the wall was 7.50 mm, 6.75 mm and 5.50 mm, respectively.

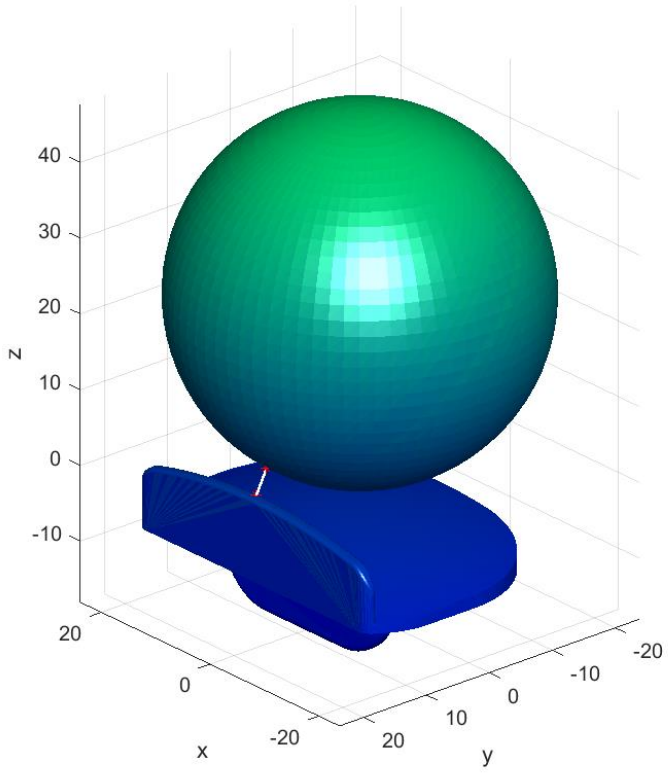


Figure 7.6. Straight line showing the smallest gap distance between the femoral sphere (representing the femoral component) and the tibial component.

The distance between the surface of the sphere of the femoral component and the top of the wall of the tibial component was related to the height of the wall (Figure 7.6). The closest distance between the sphere of the femoral component and the top of the standard wall was 1.8 mm and this occurred with the 3 mm bearing and with the bearing touching the wall. With each 1 mm increase in bearing thickness the minimum distance between the sphere and the wall increased by 1 mm. The distance between the sphere and the wall remained at 1.8 mm provided every 1 mm increase in wall height was matched by 1 mm increase in minimum bearing thickness.

As the femoral component is distracted away from the tibial component, the distance between the wall and the femoral sphere increased. For example, with a standard wall and the femoral sphere positioned as it would be with a 3 mm bearing inserted, the closest distance between the tibial wall and the femoral sphere increased from 1.8 mm to 9.4 mm as the femoral component is distracted vertically away from the tibial component from 0 mm to 8 mm.

7.4.2 Tibial dome shift results

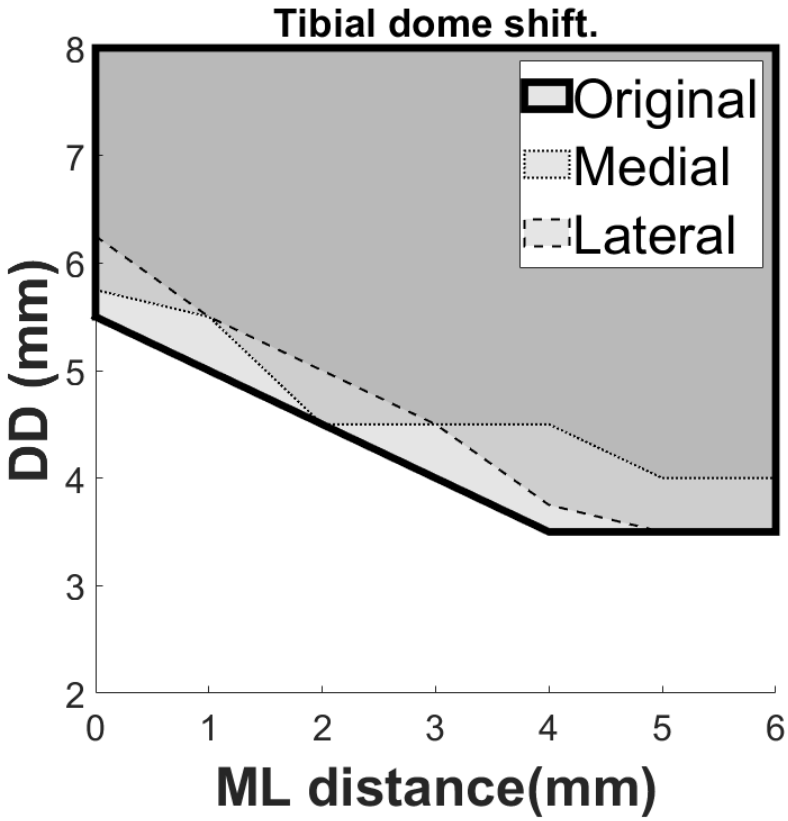


Figure 7.7. Shifting the centre of the sphere defining the tibial component either medially (towards the tibial wall) or laterally (away from the tibial wall).

Shifting the tibial dome centre of rotation 3 mm either medially or laterally increased the DD medially (Figure 7.7). Shifting medially increased the DD by 1.00 mm at 4 mm ML and 0.50 mm at 5 mm and 6 mm ML. Shifting laterally increased the DD by 0.25 mm at 0 mm, 0.75 mm between 1 mm and 3 mm ML and 0.25 mm at 4 mm ML. Between 4 mm and 6 mm ML, the DD medially is equivalent to the original implant.

7.4.3 Wider femoral component results

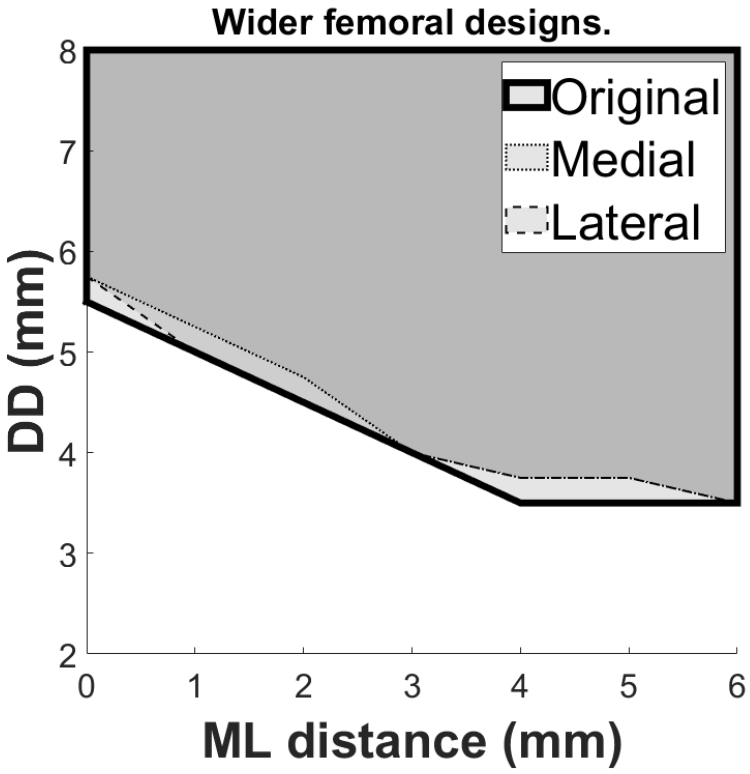


Figure 7.8. Increasing the width of the femoral component either medially (towards the tibial wall) or laterally (away from the tibial wall).

Increasing the width of the femoral component either medially or laterally had minimal effect on the DD medially with the maximum DD reduction being 0.25 mm when the femoral component was increased either medially between 0 - 3 mm ML, or either medially or laterally when the ML was between 3 - 6 mm (Figure 7.8).

7.4.4 Femoral dome shift results

Table 7.2. Tibial wall adjustments based on femoral dome shift.

Femoral sphere shift amount	Amount to adjust to reach 2 mm (mm)
3 mm medial	-1.7
3 mm lateral	+1.8

Without shifting the dome, the distance between the femoral and tibial components is 1.8 mm. Based on this value, it was aimed to retain a minimum of 2 mm between the metal components to prevent metal-on-metal collision. Shifting the femoral sphere medially by 3 mm would require the tibial wall to be decreased by 1.7 mm to maintain the 2 mm minimum distance. If the femoral sphere is shifted laterally by 3 mm, the tibial component wall could be increased by a maximum of 1.8 mm before the minimum 2 mm distance is reached.

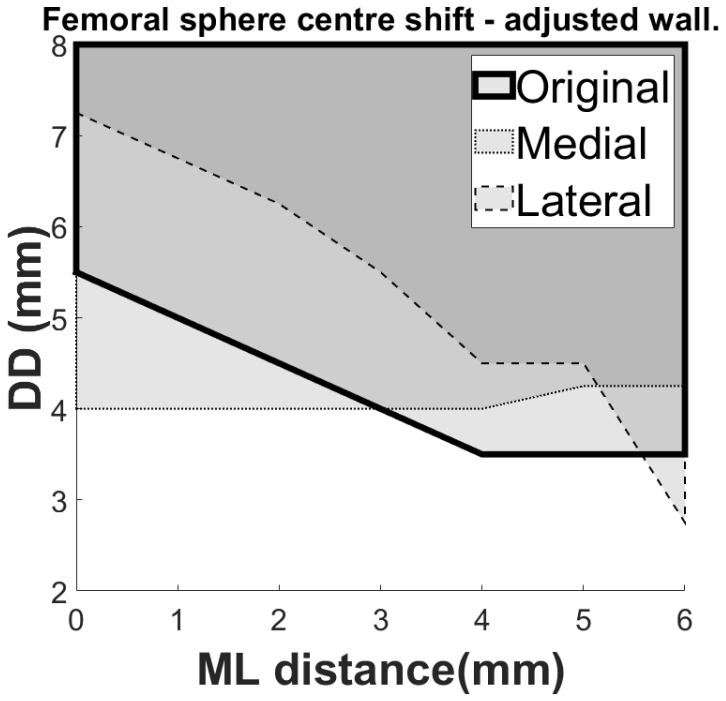


Figure 7.9. Shifting the centre of the sphere defining the femoral component either medially (towards the tibial wall) or laterally (away from the tibial wall).

Shifting the centre of the femoral sphere by 3 mm laterally and increasing the tibial wall height accordingly by 1.8 mm (Table 7.2) resulted in an increase in DD required medially by 1.75 mm if

the ML distance was 3 mm or less, and 1.00 mm if the ML distance was between 3 – 5 mm. At 6 mm ML, there is a sudden drop in performance of the implant with the DD medially dropping to 2.75 mm, which is worse than the original implant (Figure 7.9).

Shifting the centre of the femoral sphere by 3 mm medially and decreasing the tibial wall height accordingly by 1.7 mm (see Table 7.2) resulted in an initial decrease in DD required medially by 1.50 mm, however, by 3 mm ML there was no difference compared to the original implant design. Beyond 3 mm ML, shifting the centre of the femoral sphere by 3 mm medially increased the DD medially by 0.50 mm at 4 mm ML, and 0.75 mm at 5 mm and 6 mm ML (Figure 7.9).

7.4.5 Wider bearing results

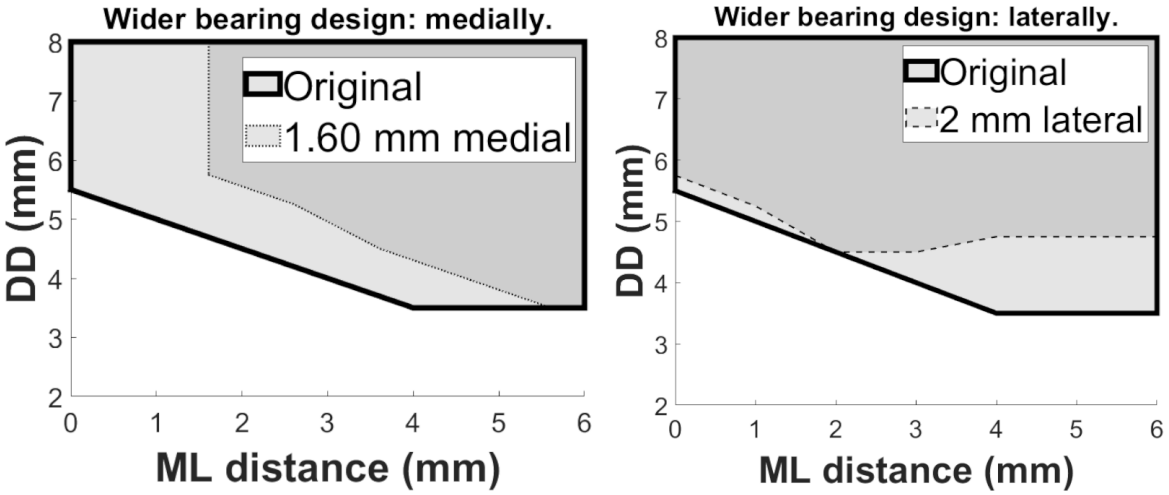


Figure 7.10. Increasing the bearing width either medially/ towards the tibial wall (left) or laterally/ away from the tibial wall (right). NOTE: Zero marks = starting point where the bearing is flush against the tibial wall if the components were unaltered. The ML distance is relative to the zero/start position.

As it was not possible to position the femoral component so that the bearing would be flush against the tibial wall using the unaltered components when the bearing is made wider medially, the starting ML is at 1.61 mm away from the tibial wall. Increasing the width of the bearing medially improved the DD by up to 1 mm if the ML was between 1.60 mm and 2.60 mm, this DD improvement drops to 0.75 mm if the ML is either 3.60 mm or 4.60 mm until ultimately at 5.60 mm and beyond, it is the same as the original implant. Increasing the width of the bearing laterally improved the DD medially by 1.25 mm if the ML was 4 mm or larger (Figure 7.10).

7.4.6 New design results

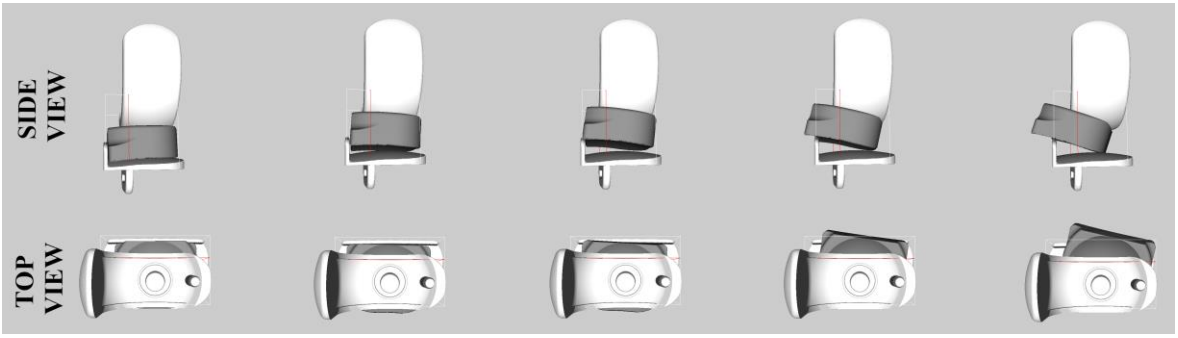


Figure 7.11. New design - one medial dislocation solution depicted in top view and side view.

Five still images taken from the animation of one representative medial dislocation with the new design is depicted in Figure 7.11. It can be seen that the bearing typically first elevates to contact the femoral component, before dislocating medially.

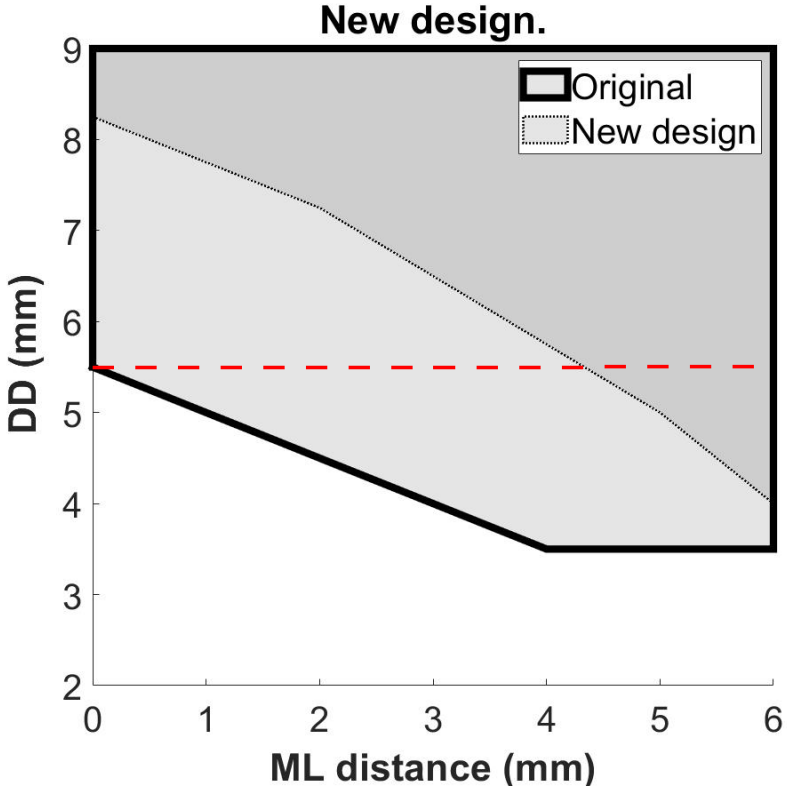


Figure 7.12. Combined design modifications to the ODL implant. NOTE: minimum bearing thickness of 4 mm. Red line shows the minimum DD for anterior/posterior dislocation at 5.5 mm.

For the new design, it was decided to increase the tibial component wall height by a total of 2.8 mm (1.8 mm adjusted wall height increase due to the femoral dome centre shift design, and an additional 1 mm as increasing the tibial wall height seemed to have a marked effect on reducing dislocation). Therefore, the new design consisted of the following design changes: 1) Shifting the centre of the femoral sphere by 3 mm laterally, 2) increasing the bearing width laterally by 2 mm, 3) increasing the tibial component wall height by a total of 2.8 mm and 4) using a minimum bearing thickness of 4 mm. The new design successfully reduced the medial mobile bearing dislocation risk (Figure 7.12). When the bearing is flush with the tibial wall, or positioned at 2 mm, 4 mm or 6 mm ML, the DD medially is increased to >8.00 mm, 7.25 mm, 5.75 mm and 4.00 mm respectively (Figure 7.12). As the goal was to make the DD medially match the anterior/posterior DD, it is evident that this condition is only satisfied if the ML is 4 mm or less (Figure 7.12).

7.4.7 Validation using the mechanical rig

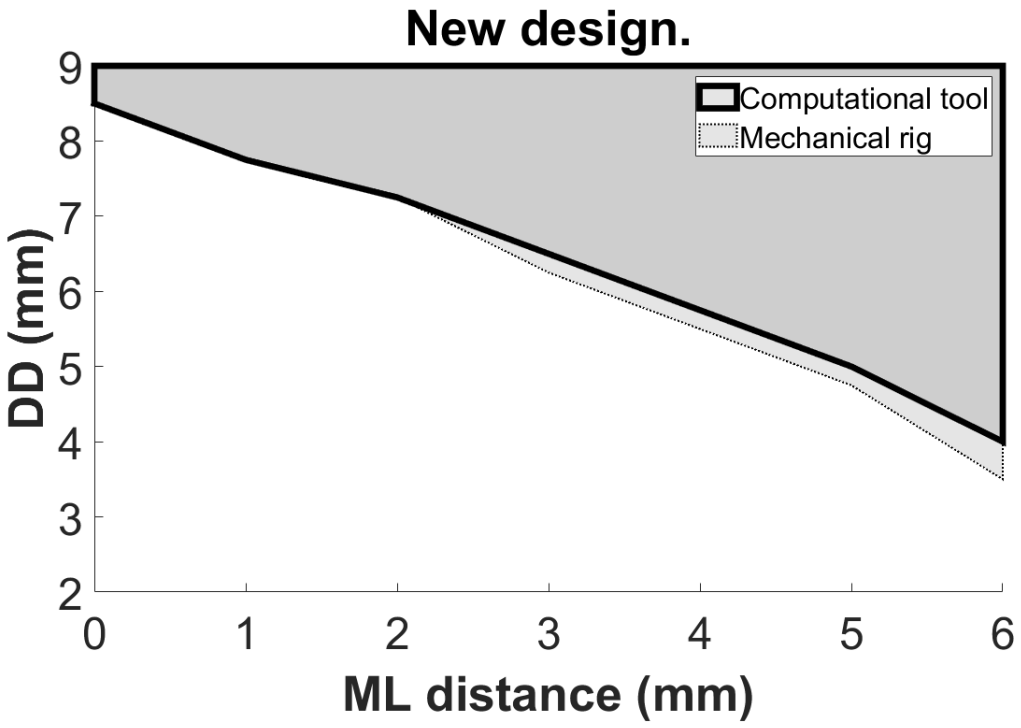


Figure 7.13. Validation of the results for the new design of the ODL implant comparing results obtained using the mechanical rig and the dislocation analysis tool.

The results between the computational dislocation analysis tool and the modified mechanical rig (Figure 7.13) show identical results at 0 mm ML distance. By 6 mm ML, the mechanical rig required 0.50 mm less DD than the dislocation analysis tool for a dislocation (3.50 mm versus 4.00 mm).

7.5 Discussion

The risk of bearing dislocation is related to the amount of entrapment of the mobile bearing, which can be quantified by the minimum amount of distraction required for a bearing dislocation to occur (DD). The larger the DD, the more difficult it is for dislocations to occur and therefore the lower the risk of bearing dislocation. This chapter aimed to modify the design of the ODL implant to reduce the medial bearing dislocation risk when compared with the original components for a large number of clinically relevant positions of the femoral and tibial components. Since anterior and posterior dislocations are rare, occurring in less than 1% of cases [6], the goal was to modify the design of the implant to increase the DD for a medial dislocation to match the minimum anterior/posterior DD of 5.5 mm, so that the overall risk of bearing dislocation would probably be acceptable.

In this chapter, for each component in the ODL implant, various implant design modifications that could reduce the medial dislocation risk were considered. While shifting the femoral sphere centre 3 mm medially adversely affected the medial dislocation risk, shifting the femoral sphere centre 3 mm laterally reduced the medial dislocation risk. Other design changes that reduced the medial dislocation risk included increasing the mobile bearing width 2 mm laterally, or increasing the femoral component width either 2 mm medially or laterally. It is not recommended to increase the mobile bearing width more than 2 mm laterally, as this may result in lateral bearing overhang. While increasing the width of the mobile bearing medially reduces the dislocation risk, the surgeon is required to implant the femoral component further laterally relative to the tibial component which reduces the acceptable ML range to less than 4 mm. While shifting the tibial dome either medially or laterally also reduced the medial dislocation risk, this design change was not advantageous since the benefit was relatively small, and more critically, it would require the use of a thicker tibial component which would in turn require a lower tibial cut (and more bone resection), therefore this

design modification was abandoned. The "new design" combined the individual design changes to each component that successfully reduced the medial dislocation risk. It consisted of the following design changes: shifting the femoral sphere 3 mm laterally, increasing the bearing width laterally by 2 mm and increasing the tibial component wall height by a total of 2.8 mm.

Shifting the femoral sphere 3 mm laterally enabled the wall height to be increased by 1.8 mm. While this design change increased the DD medially by 1.75 mm for ML ≤3 mm, and 1 mm if the ML distance was between 3 – 5 mm, the DD medially suddenly drops to 2.75 mm at 6 mm ML, which was worse than the original implant. To counter this, the bearing width was increased laterally. To further try and address this, it was also decided to increase the tibial component wall height by an additional 1 mm to increase the medial DD by 1.00 mm. At operation the surgeon aims to align the femoral and tibial components so that they are approximately parallel with the knee in flexion. During knee movements such as flexion/extension and rotation, the components rotate relative to each other so part of the femoral component could be above or potentially hit the wall. With the current implant design the minimum distance between the femoral sphere and the top of the wall, which occurs when the thinnest bearing, which is nominally 3 mm, touches the wall, is 1.8 mm. The linear wear rate of the ODL is very low [188] and no cases where the femoral component has hit the tibial wall have been reported in the literature, so this distance between the components is safe. Therefore, to prevent metal-on-metal collision between the femoral and tibial components, increasing the tibial wall height by an additional 1 mm beyond the 1.8 mm increase in wall height after the femoral dome centre was shifted 3 mm laterally, required the use of a minimum bearing that is correspondingly 1 mm thicker. Clinically, this would require an additional 1 mm more tibial bone resected. This would be unlikely to cause any problems. It would require a small amount of the ilio-tibial tract to be released from the upper part of Gerdy's tubercle, but no ligament release. The situation is very different from that on the medial side of the knee where a deep resection is likely to cause problems as it would require release of part of the deep medial collateral ligament, which might lead to over correction. The use of a 4 mm bearing rather than a 3 mm bearing would have a small additional advantage as it would increase the DD by about 0.10 mm [189]. Therefore, the minimum bearing thickness for the new

design was 4 mm and that the wall height was 2.8 mm higher than the original tibial wall height, but remained 2 mm away from the femoral sphere. These combined changes successfully countered the sudden drop in DD between 5 mm and 6 mm.

As the goal was to make the DD medially of the new design match the minimum anterior/posterior DD of the original ODL implant (5.5 mm), the ML needs to be at maximum 4 mm away from the tibial wall when the knee is at 90° flexion because this is when the laxity of the Lateral Collateral Ligament (LCL) is at maximum [29] and when the bearing is furthest from the tibial wall. Therefore, other design changes that may increase the DD medially particularly between 4 – 6 mm ML are required. While increasing the wall height by an additional 1 mm seems like a simple solution, clinically, this would require an additional 1 mm more bone resected. Another solution could be to use a non-metal material to increase the tibial component wall height but mechanically, this seems challenging.

In addition to modifying the design of the ODL implant components, the surgical technique may also be altered to reduce the dislocation risk. The further the ML distance between the mobile bearing and the tibial component wall, the larger the dislocation risk, therefore, if surgeons were to use the new implant design, the position of the bearing should be adjusted so that it does not jam against the wall or have >4 mm of ML distance from the wall if they aim to reduce the risk of medial dislocation. Practically, this means that prior to preparing the keel slot, surgeons should do a trial reduction and assess the bearing position relative to the tibial wall in all positions. Since the ODL implant consists of three components (femoral component, tibial component and the polyethylene mobile bearing) and multiple steps to implant each of the components correctly, the final mediolateral distance between the metal femoral and tibial components is effectively a cumulative summation of individual distances from the steps within the surgical technique that can affect the mediolateral placement of the femoral and tibial components. Notable steps in the surgical procedure which can affect the final mediolateral translation distance are: the initial tibial plateau cut with internal rotation, femoral component placement (mediolateral placement, varus/valgus angle and the relative determination of the centre of the femur as a reference point) and appropriate gap balancing.

Other relative positions of the femoral and tibial components which may affect dislocation include rotating the tibial component, or tilting the femoral component either towards or away from the tibial wall. Since, in Chapter 3, it was shown that rotating the tibial component did not affect the dislocation risk, in this study, tibial rotation was ignored. In this study, the effect of increasing the femoral width by 2 mm medially was studied, which has the same effect as tilting the femoral component towards the tibial wall by approximately 0.5° towards the tibial component wall. Doing so added approximately 1 mm DD; further research could look at the effect of tilting the femoral component in more detail.

The main limitation of the study is that it assumes that dislocation is solely caused by implant geometry. Other factors which might push the bearing out of place once the lateral compartment is distracted, and thus contribute to bearing dislocation, have not been assessed. Further cadaveric and clinical testing is required to confirm that the new proposed design does in fact reduce the risk of medial dislocation, without introducing any new and unexpected complications.

7.6 Conclusions

This chapter assessed design modifications to the femoral component, tibial component and the mobile bearing in the ODL implant and found that for all designs, the dislocation risk is largest when the ML distance between the mobile bearing and the tibial wall is 4 mm or more. Therefore it is strongly recommended that the surgical technique be altered to ensure that the ML is 4 mm or less. However, simply changing the surgical technique does not sufficiently reduce the risk of medial dislocation risk to an acceptable level. Therefore, the implant design also needs to be altered.

If the surgical technique can be altered so that the bearing never hits the wall, the simplest design change would be to increase the tibial component wall height by at least 2 mm of the current design. However, further work would need to identify the maximum increase in bearing thickness that would not compromise the bone because for each 1 mm increase in wall height, an additional 1 mm of bone needs to be removed. Otherwise if this is not possible, the components can be altered to include the following changes: shifting the femoral sphere centre laterally by 3 mm, increasing the width of the

mobile bearing laterally by 2 mm and increasing the tibial component wall height by 2.8 mm. This would reduce the medial mobile bearing dislocation risk to an acceptable risk. It has been demonstrated in this chapter that combining these three modifications successfully increases the DD to 5.75 mm which is larger than the minimum anterior posterior dislocation DD of 5.50 mm providing the surgeon implants the components so the ML distance 4 mm or less.

7.7 Acknowledgements

Mr. Thomas W. Hamilton is thanked for his assistance in identifying the steps within the surgical technique that contribute towards determining the final relative ML position.

Chapter 8 THESIS CONCLUSIONS AND SUGGESTIONS FOR FUTURE WORK

8.1 Thesis conclusions

The most common complication with the ODL UKR implant (Zimmer Biomet, Bridgend, UK) is dislocation of the mobile bearing [7, 10, 20, 119]. A recent study using NJR data found that bearing dislocations accounted for 30% of the revisions using the ODL implant [119]. The exact bearing dislocation rate seems to vary considerably. Some studies have reported a bearing dislocation rate of 6% using the ODL implant [40] while others have reported dislocation rates as low as 2% [39, 41]. Using a meta-analysis in Chapter 2, the cumulative bearing dislocation rate was found to be 3.7% using the current ODL implant design, which is similar to the most recent dislocation rate reported by the designer surgeon group of 4% [6] but slightly higher than the dislocation rate calculated using NJR data of 2.3% [119]. Nevertheless, bearing dislocation remains a problem in the lateral knee, with the bearing dislocation rate being up to 10 times higher in the lateral knee as compared with the medial knee (0.58% [147]).

The most likely reason for the higher dislocation rate laterally is because when the knee is flexed, the lateral ligaments are lax whereas medially the ligaments are tight [164], allowing greater distraction in the lateral versus the medial knee. While clinically, dislocations are possible medially, anteriorly and posteriorly, in Chapter 2 it was found that most dislocations occur medially with the bearing dislocating into the knee joint. It was also found that dislocations tended to occur early, on average 16 months post-operatively, which is probably why the ODL can suffer from poor early survivorship [40] compared to fixed bearing designs [42]. Finally, while the exact cause for dislocation was unknown, most dislocations have been reported either following trauma [39, 41, 146], or during squatting [162], when the knee is flexed.

To further investigate the relationship between knee flexion and dislocation, in Chapter 3, using a modified rig, clear definitions for dislocation and the dislocation configuration yielding the lowest DD value, the influence of tibial component rotation, posterior bearing overhang and mediolateral

translation on bearing dislocation were studied. Using the rig, the components were positioned in a range of physiologically relevant positions that the components may assume when the knee is flexed. It was found that vertical distraction was required for dislocation and the amount of Distraction to Dislocation (DD) varied depending on the mediolateral (ML) position. Testing found that lateral dislocations required the least DD, however, clinically, lateral dislocations do not occur probably because the lateral compartment has a curtain of ligaments and other soft tissues which act as a physical barrier preventing lateral dislocations. The DD medially was lower than the DD anteriorly/posteriorly (5.75 mm versus 6.00 mm respectively). Physiologically, the lateral compartment of the knee can distract on average 6.7±1.9 mm following application of a varus stress to a knee joint in 90° of knee flexion [32] which is just enough for an anterior or posterior dislocation, whereas it is easily enough for a medial dislocation especially the further the bearing is positioned away from the tibial wall (2.25 mm at 6.00 mm ML). This finding represents one possible reason why clinically, medial dislocations are more common than anterior/posterior dislocations. While internal or external rotation were not found to affect the dislocation risk, one study found that externally rotating the tibial component can lead to negative post-operative clinical outcomes [177], so this should be avoided. Posterior bearing overhang as a result of flexion was also found not to affect the bearing dislocation risk, except when the flexion angle was 140° which reduced the DD for medial dislocation by 1 mm, however, this is an extreme amount of flexion that is both uncommon [158] and unlikely. Since tibial rotation and posterior mobile bearing overhang had little effect on the dislocation risk, subsequent studies focused on testing the DD at different ML positions. Therefore, based on the findings in this thesis, the results suggest that the maximum dislocation risk likely occurs when there is maximal distraction between the femur and tibia and when the bearing is furthest away from the tibial wall. Given that when the knee is flexed at 90°, the Lateral Collateral Ligament (LCL) reaches maximal laxity [29] (and therefore when maximal distraction in the lateral knee is possible), and the bearing is furthest from the tibial wall, it is likely that the risk of mobile bearing dislocation is highest when the knee is flexed at 90°.

One way to reduce the dislocation rates seen clinically, is to alter the design of the implant. In 2006, the design of the tibial component was altered from a flat tibial surface to a domed tibial surface to reduce the risk of bearing dislocations [159]. The meta-analysis conducted in Chapter 1 found that this design change reduced the dislocation rate from 17% to 3.7% with most recent studies reporting a similar dislocation rate of 2 - 4% [6-8, 39, 41, 119] including a study using NJR data which found a 2.3% dislocation rate using the ODL implant [119]. While the overall incidence of dislocation has reduced over time, it was demonstrated in Chapter 5, that changing the design of the tibial component from a flat to a domed tibial surface had a greater impact at reducing the rate of anterior/posterior dislocations and the reason for this is because the design modification increased the entrapment of the bearing anteriorly/posteriorly, with minimal impact of the medial entrapment of the mobile bearing. As a result, this reduced the anterior/posterior dislocation risk more than the medial dislocation risk. Taken together, Chapter 1 and Chapter 5 demonstrate that the bearing dislocation risk is amenable to manipulation by changing the design of the implant, however, that dislocations of the bearing medially remain a problem. Therefore, Chapter 6 explored other design changes that could be made to the ODL implant design to further reduce the medial dislocation risk to an acceptable level, without compromising the current implant functionality and performance. The features that could be altered are presented in Table 6.2 and the selected design modifications are segregated by component: tibial component (Table 6.4), femoral component (Table 6.5) and the mobile bearing (Table 6.6).

To test the effectiveness of the implant design modifications against bearing dislocation, one option was to use the custom-built mechanical rig described in Chapter 3. However as testing with the rig was inefficient and therefore impractical, a novel dislocation analysis tool was developed and validated and this process was described in Chapter 4. The dislocation analysis tool automates the bearing motion to find dislocations given any relative setup of the femoral and tibial components. The output of the tool is the DD either anteriorly, posteriorly, medially or laterally. By comparing the minimum DD for each setup, this allows quantification of the mobile bearing dislocation risk: the higher the minimum DD, the more difficult dislocations are and vice versa. In addition, since the

tool automates bearing motion, it gives unique insight into the possible pathways that the bearing may take from a non-dislocated position to a dislocated position given any relative setup of the metallic femoral and tibial ODL components.

In this thesis, the dislocation analysis tool was used to test and compare new implants with the original ODL implant design. The goal was to reduce the DD medially to match that of DD anteriorly/posteriorly, because anterior and posterior dislocations occur at a clinically acceptable rate. Matching the medial DD to the anterior and posterior DD would result in the medial dislocation risk being equivalent to the anterior and posterior dislocation risk, and thereby, it is hoped that this can result in an acceptable medial dislocation rate. In Chapter 7, it was demonstrated that to facilitate efforts to reduce the risk of bearing dislocation medially, it is strongly recommended that the surgical technique needs to be altered to ensure that the ML is 4 mm or less. If the surgical technique can be altered so that the bearing never hits the wall, the simplest design change would be to increase the tibial component wall height, however, since each millimetre of tibial wall height increase requires a millimetre of tibial bone excision to avoid metal-on-metal collision, the maximal increase that does not compromise the bone integrity needs to be investigated. Another alternative would be to increase the wall height using a non-metal material so that metal-on-metal collision is not a concern. If these options are exhausted, relatively simple modifications can be made to each of the ODL implant components: increase tibial component wall height by 2.8 mm, shift the femoral dome centre 3 mm laterally and increase the bearing thickness 2 mm laterally. This thesis also suggests, in line with previous work [10], that if surgeons want to best reduce the risk of medial dislocations, they should aim to implant the ODL implant bearing as close as possible to the tibial component wall without hitting it. If the design modifications are made as suggested, the surgeon should aim to implant the bearing no more than 4 mm ML away from the tibial component wall in order to match the AP dislocation risk.

8.1 Overall thesis limitations

Isolated lateral unicompartmental disease is relatively rare, accounting for 10% of all arthritic knees [4], therefore, few lateral UKR are implanted and few publications with limited data were available for assessment. In Chapter 2, a dislocation rate was deduced based on the available studies however, confidence in the dislocation rate estimate reported is limited due to the limited number of studies available, so the true dislocation rate may be different from the estimate provided in this thesis. Dislocations were studied using either a mechanical rig (Chapter 3) or a computational dislocation analysis tool (Chapter 4). Testing dislocations with the mechanical rig was time consuming and subject to user error – variability in results could occur due to tester fatigue, subjective judgement of force applied to achieve dislocation. The dislocation analysis tool effectively carried out the same analysis as the mechanical rig but overall, it was more efficient, more versatile (allowing different design models to be tested) and less prone to user error than the mechanical rig. However, using either the computational dislocation analysis tool or the mechanical rig, many factors that might cause the bearing to dislocate once the lateral compartment is distracted, e.g. the effect of the synovial fluid and soft tissues in the knee joint on dislocation, could not be assessed. Nevertheless, the results obtained using either tool which measured the minimum DD of the femoral component away from the tibial component for various relative configuration setups of the metal components seem to be valid as it explains why clinically, medial dislocation is most common. In Chapter 5 it was found that while previously changing the design of the tibial component from a flat tibial plateau to a domed tibial plateau reduced the overall dislocation rate, it was shown that anterior/posterior dislocations were reduced more than medial dislocations, however, these findings were calculated based on limited data and in some cases, informed assumptions were made regarding the direction of dislocation in instances where these were not explicitly reported. Nevertheless, the data did demonstrate that overall, medial dislocations are more common than anterior/posterior dislocations. Therefore, to address the problem of medial dislocation, this thesis focused on altering the geometry of the implant to prevent the bearing from escaping medially. It is very likely that the problem of bearing dislocation is a multifactorial one involving patient selection, surgeon experience (high versus low volume surgeons), surgeon type (designer versus non-designer) implant design, surgical

technique and surgical instrumentation. While some recommendations regarding changing the surgical technique to reduce dislocation risk were made, this was not explicitly explored. However, based on evidence that changing the design of the tibial plateau from a flat to a domed design reduced the dislocation rate clinically, it appears acceptable to assume that further alterations to the implant geometry could further reduce the clinical dislocation rate, however further testing is required to confirm whether the proposed design changes to the implant sufficiently reduce the clinical dislocation rate to an acceptable rate. The experiments setup and the results presented in this thesis do not take into account the native knee structures such as the synovial fluid, the soft tissues in the knee joint such as the joint capsule, the ligaments inside the knee joint (LCL, ACL, MCL) and the patellar bone, which could all have an effect on the dislocation risk and therefore the dislocation rate. The configurations chosen for testing in this thesis were based on static relative positions that the knee could assume throughout the knee flexion cycle, however, the knee is a dynamic joint involving many forces, which were not considered in these assessments. Further research would also explore the specific joint forces that correspond to the various implant configurations that have been presented, as these are likely to play a role in informing how dislocations occur and therefore, how best to prevent these dislocations, or assess dislocation in a dynamic knee joint. While multiple criteria were used to assess the feasibility of the design modifications proposed in this thesis, financial constraints were not considered, which could be a large hurdle to overcome to ensure translation of these ideas into clinical practice. Further research would explore the pathway to translation in order to bring these concepts to patients.

8.2 Future work

To build on the work presented in this thesis, further research should focus on methods that can alter the surgical technique in a simple and safe manner to ensure that the ML distance between the mobile bearing and the tibial wall is 4 mm or less. However, since changing the surgical technique does not sufficiently reduce the risk of medial dislocation risk, further research would focus on altering the implant design. If the surgical technique can be altered so that the bearing never hits the wall, the

first implant design change would investigate the simplest design change which would be to increase the tibial component wall height by at least 2 mm of the current design. Further work would need to identify the maximum increase in bearing thickness that would not compromise the bone because for each 1 mm increase in wall height, an additional 1 mm of bone needs to be removed. If this is not possible to alter the surgical technique so that the bearing never hits the wall, the components can be altered to include the following changes: shifting the femoral sphere centre laterally by 3 mm, increasing the width of the mobile bearing laterally by 2 mm and increasing the tibial component wall height by 2.8 mm. Further research could also investigate other design changes that could remove the conditional requirement for surgeons to implant the bearing a distance of 4 mm or less from the tibial component wall could also be explored to allow surgeons to implant the device up to 6 mm away from the tibial component. This would provide a more flexible target for surgeons, ensuring greater confidence that the dislocation risk to fall within the clinically acceptable range.

Once the design of the implant has been altered, further research is required to ensure that the new implant design satisfies the user needs as listed in Table 6.1. Most importantly, further cadaveric and clinical testing will verify that the design modifications do not compromise the ODL implant performance or introduce any new adverse effects. Finally, if all of this can be achieved, long term data collection is required to enable quantification and analysis of the clinical medial dislocation rate to determine whether these changes have been successful or not.

Regarding the dislocation analysis tool developed in this thesis, the tool can be used to study the path that the bearing takes to dislocate, and this information can be used to inform design changes which aim to impede this path to reduce dislocation risk. If individual ligament laxities can be measured, future work could also explore developing the tool further for use diagnostically, wherein the specific device, and relative position of the components (including the amount of distraction in the knee) is provided to the tool, and the tool provides a numeric value for the risk of dislocation. Overtime, after collecting sufficient information, a normal distribution can be obtained which may facilitate to provide greater context and meaning to the output risk value. A logical follow on study to build on the work in this thesis could assess the contact points and kinematics of a mobile bearing through a

range of knee flexion angles (perhaps using dynamic RSA) and capture the relative positions of the femoral and tibial components (perhaps using imaging techniques) in a patient who has sustained a clinical dislocation. These same relative component positions of the femoral and tibial components could then be modelled in the dislocation analysis tool and for each configuration setup, the dislocation could then be assessed with a specific comparison of the ODL implant versus the new design proposed in Chapter 7. This study would give valuable insight into the possible physiological positions that the knee would have been in when the patient sustained a dislocation and whether the new design could reduce the risk of bearing dislocation.

8.3 Conclusion

Like all innovation ideas that are intended for translation into clinical practice, many factors have been considered to determine the feasibility of the proposed design changes. A large portion of this engineering design project has been weighing up the innovation and science alongside the clinical interests/demands. Further, additional considerations like the current scientific advancement may also play a role in the overall determination of design feasibility. Therefore, while some ideas presented in this thesis may have been rejected, they may still hold some innovative or scientific relevance in the future. For instance, the concept of a “deformable bearing”, which may look like a bearing sliced along the centre horizontally and sandwiching a spring that compresses under knee loading and expands under tension when the knee is offloaded, may one day be feasible if a suitable material is developed.

Currently, revision of the mobile tibia to the FLO is probably the best and most commonly used treatment [40, 42, 190], however, it is hoped that the work in this thesis demonstrates that there remains potential to further work to improve the design of the ODL implant and that if the medial dislocation risk can be reduced, then the mobile bearing would be a favourable implant particularly in the ever growing younger lateral OA patients who may benefit from the many additional advantages that mobile bearing designs like the ODL implant have to offer.

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Appendices

APPENDIX A: MATLAB COMPUTATIONAL TOOL TO ASSESS BEARING DISLOCATION

The following depicts the method, results and conclusions obtained from the initial dislocation analysis code developed in MATLAB.

Method

CAD models of the latest model domed ODL femoral component and the tibial component were obtained from the manufacturer, Biomet (Swindon, UK). The files were exported to STL file format using SOLIDWORKS 2017 (Dessault Systemes SOLIDWORKS Corp., Waltham, MA, USA). The STL files were imported into MATLAB (version R2018b, MathWorks Inc., MA, USA) and converted into point clouds. To control the relative component positions, each point cloud was rotated and translated using rotation matrices. The components were positioned with the bearing congruent with the femoral and tibial components. The lowest point of the femoral component was vertically aligned with lowest point of the bearing superior surface. Similarly, the highest point of the domed surface of the tibial component was vertically aligned with highest point of the bearing inferior surface. Following positioning of the components, the tibial component was converted into a boundary surface using the MATLAB inbuilt alphaShape function.

To determine dislocations of the mobile bearing, the computational model was built in MATLAB. Based on the results obtained in Chapter 3, tibial rotation and knee flexion were ignored. A MATLAB code was written assuming full congruency of the mobile bearing with the femoral component at all times. The tibial component was converted using the alphaShape function into a boundary surface. Rotation of the bearing was enabled in the transverse plane first. Following this, sagittal plane rotation of the mobile bearing was enabled radially about the centre of rotation of the femoral component sphere, which is located at the top of the femoral peg. Finally, the bearing was allowed to swing through a range of angles in the coronal plane. At each coronal plane rotation angle, the

MATLAB inShape function was used to assess whether any of the points of the bearing point cloud, fell inside the converted alphaShape boundary of the tibial component. If any bearing point was found to be inside of the tibial boundary surface, the code would record a dislocation. If across all the coronal plane rotation angles tested, any one rotation angle showed overlap of the bearing component with the tibial wall, a dislocation was recorded. Testing parameters are listed in Appendix A Table 1.

Appendix A Table 1. MATLAB inShape dislocation testing parameters.

Testing parameter (unit of measurement)	Testing range (increments)
Vertical distraction (mm)	0-8 (0.25)
Mediolateral translation (mm)	0-4 (1)
Transverse plane bearing rotation angles	30;15;10;0;-10;-15;-30
Sagittal plane bearing rotation angle range (°)	0;1;2;3;4-20 (2)
Coronal plane bearing rotation angles (°)†	-40;-30;-20;-18;-16

†Negative indicates rotation of the mobile bearing towards the tibial wall

In order to validate the MATLAB code, the results for the manual testing were plotted in the same graph against the results obtained using the computation code. A Pearson's correlation coefficient (r) was calculated using the following formula:

$$r = \frac{\sum (x_i - \bar{x})(y_i - \bar{y})}{\sqrt{\sum (x_i - \bar{x})^2 \sum (y_i - \bar{y})^2}}$$

Where:

$r = correlation\ coefficient$

$x_i = x\ values\ in\ a\ sample$

$\bar{x} = mean\ of\ the\ x\ values\ in\ the\ sample$

$y_i = y\ values\ in\ a\ sample$

$\bar{y} = mean\ of\ the\ y\ values\ in\ the\ sample$

However, from this plot, it was not clear how different or similar manual testing using the mechanical rig was when compared to the computational testing results. Therefore, a plot of the results obtained

from manual testing (x-axis) were plotted against the results obtained from computational testing (y-axis). A linear regression model or a linear line of best fit was modelled to this plot and the coefficient of determination (r^2) was calculated to assess the proportionate variation in y values with respect to x values in a linear regression model.

Additionally, a Bland Altmann [191] analysis was conducted and a Bland Altmann plot generated demonstrating average vertical distraction to dislocation (mm) measured by manual testing and computational testing (x-axis), versus difference in vertical distraction to dislocation (mm) (computational - manual) (y-axis). The average and difference for using manual testing or computational testing for each measurement point was calculated for the dataset using the formulae below.

$$Mean\ of\ each\ data\ point = \frac{\mathrm{x}_1 + x_2}{2}$$

$$Difference\ between\ each\ data\ point = \ |x_1 - x_2|$$

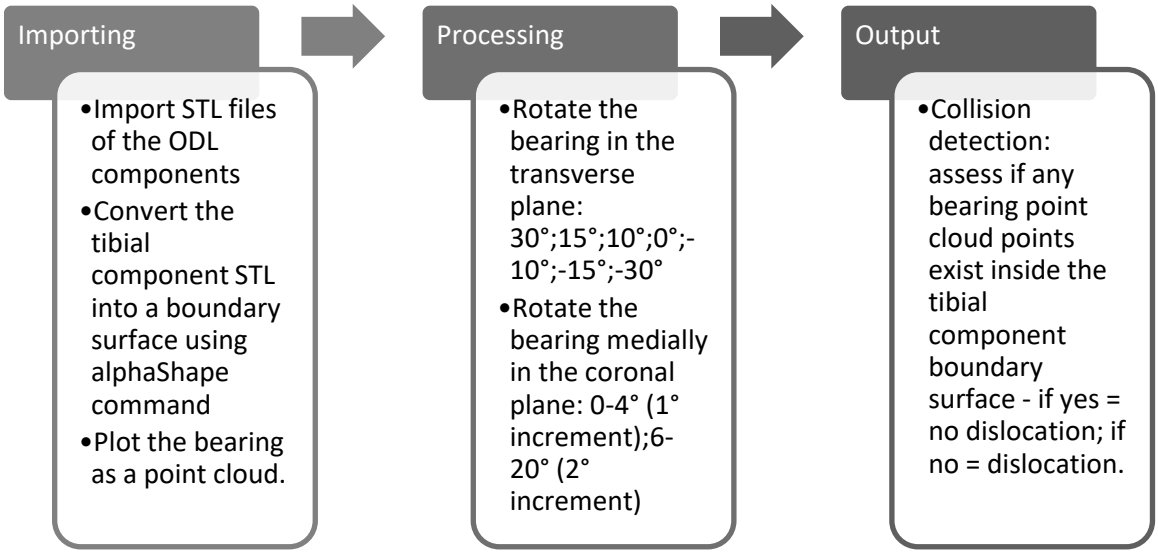
Following this, the mean and the standard deviation for the difference between each data point was calculated.

$$Mean\ of\ difference\ values = \frac{\Sigma(Difference\ between\ each\ data\ point)}{n}$$

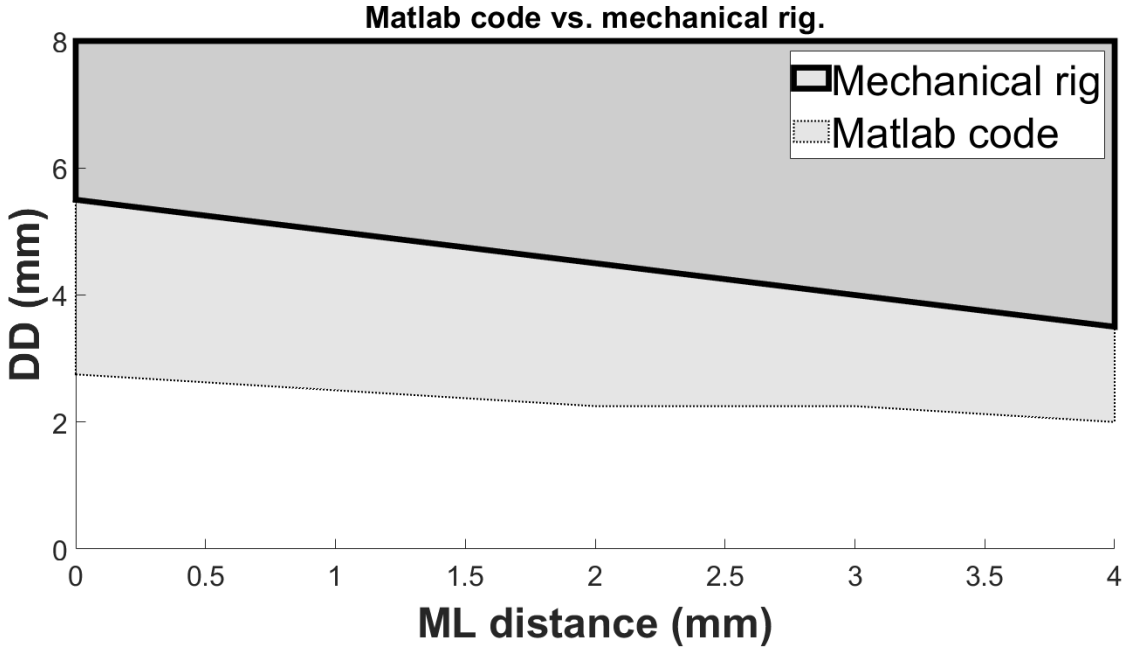
Where n is the total number of data points measured.

$$Standard\ deviation\ of\ difference\ values = \sqrt{\frac{\Sigma(x - \bar{x})^2}{n}}$$

Results



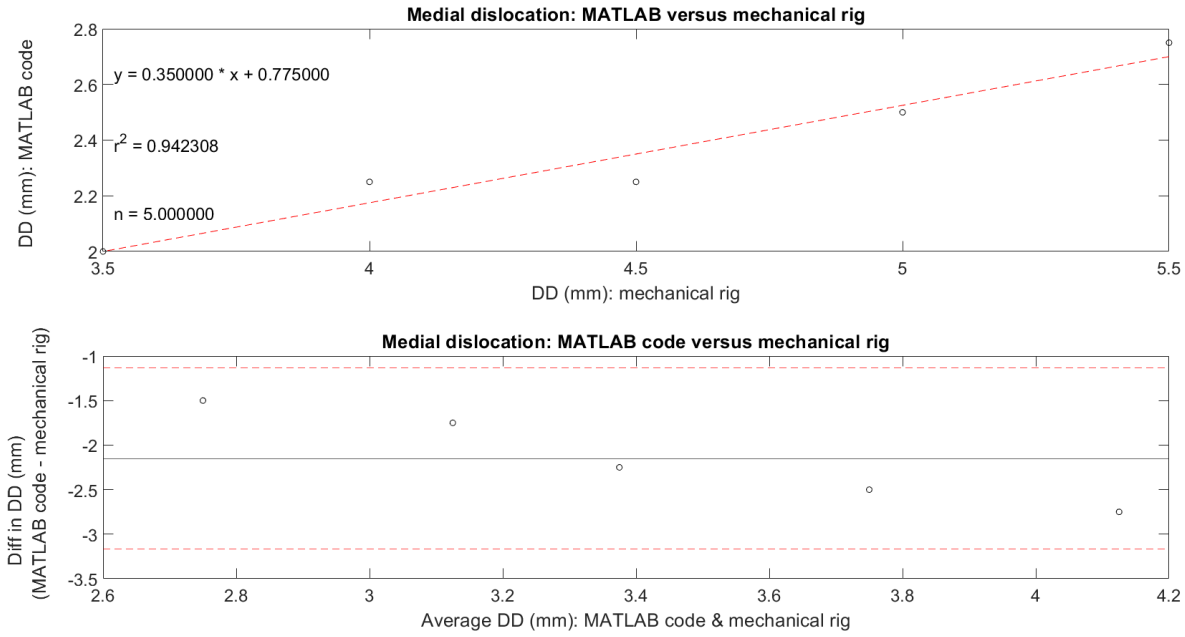
Appendix Figure 1. The code workflow.



Appendix Figure 2. Plot showing the results obtained using obtained using the MATLAB code or the mechanical rig.

Results obtained using the MATLAB code followed the same trend as the results obtained using the mechanical rig, however, with a slightly different gradient (Appendix Figure 2). Both the dislocation

results obtained using the MATLAB code and the mechanical rig showed a negative trend, indicating that as mediolateral translation increased, dislocation became easier. At 0 mm, 2 mm and 4 mm ML, the DD medially using the mechanical rig were 5.5 mm, 4.5 mm and 3.5 mm while using the MATLAB code were 2.75 mm, 2.25 mm and 2 mm.

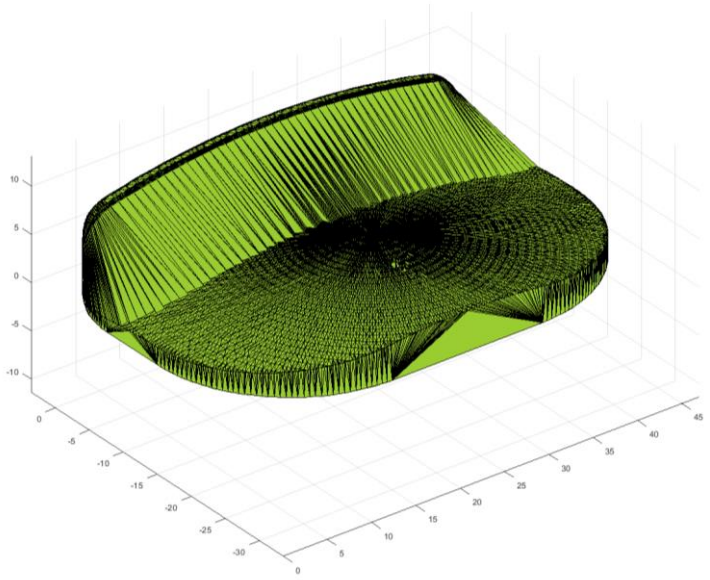


Appendix Figure 3. Bland Altmann comparison of bearing dislocation results using either the MATLAB code and the mechanical rig.

When comparing the results obtained using either the MATLAB code or the mechanical rig, the r^2 value was 0.94. The Bland Altmann results (Appendix Figure 3) indicated that the MATLAB code results required on average 2.15 mm less DD when compared with the mechanical rig.

Discussion

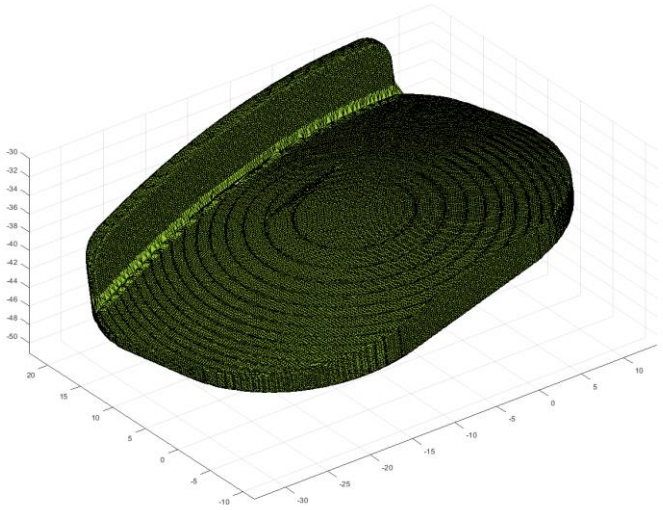
In this study a successful MATLAB dislocation model was developed which improved on the design of the mechanical rig by removing user dependency, reducing the total testing time and allowing any new geometry to be tested. The alphaShape function used to model the tibial component was not adept at dealing with perpendicular junctions (e.g. the junction between the tibial wall and the domed surface), adding therefore a skirt connecting the tibial wall and the domed surface (Appendix Figure 4).



Appendix Figure 4. Skirt created after applying alphaShape function in MATLAB.

As this was no longer representative of the geometry of the tibial component, it was decided to split the tibial component into two sub-portions: the tibial wall and the domed portion. Therefore, separate boundary surfaces were created for the tibial wall and the domed portion for analysis.

Despite separating, the whole surface remained representative of the tibial component (Appendix Figure 5). This deviation did not affect the results, since testing was split into analysing dislocations with the tibial wall and the domed portion.



Appendix Figure 5. Final alphaShape of tibial component used for testing

In our testing, a dislocation was recorded when any point of the bearing component point cloud fell within the boundary surface of the tibial component wall. However, there may be some scenarios where a slight tweak in the 3D orientation of the mobile bearing, would prevent the bearing from being dislocated, but since the components were not active/smart, these slight nuanced motions were not enabled. Since not much is known about what causes dislocation, defining dislocation was a challenge: the exact range of positions/ 3D orientations that the bearing may assume when it is dislocated may vary.

The MATLAB dislocation analysis tool assumed that the bearing would remain in full congruency with the femoral component irrespective of the amount of vertical distraction or mediolateral translation in the knee. Initially, this assumption seemed intuitively correct – it would be more likely for the bearing to dislocate, if the bearing found a way to maximise on the gap between the femoral and tibial components, in order to dislocate. However, it is currently unknown exactly how the bearing dislocates, and therefore whether the bearing is always in full congruency with the femoral component before dislocating. Further, using the MATLAB dislocation model, the path that the bearing takes in order for a dislocation to occur was defined to be radial in three sequential planes: transverse, sagittal and then coronal, and always in this order. However, from manual testing using the mechanical rig, it is clear that bearing dislocation is complex often involving a complex range of motions that sometimes alternated between rotation, translation and a combination of these to navigate from between the femoral and tibial components.

However, the biggest issue using the MATLAB code was that many assumptions regarding the motion of the bearing to achieve dislocation were made which resulted in a different trend regarding the effect of ML translation on medial bearing dislocation and deviations of more than 1 mm in the final results when compared with the mechanical rig. Additionally, the MATLAB code had not been optimised to interrogate the solution space in a computationally efficient way and it was not practical to improve on this design especially in comparison with the approach of using path planning algorithms. Dislocation of the mobile bearing computationally required on average 2.15 mm less vertical distraction as compared with the mechanical rig, which is far more than the acceptable

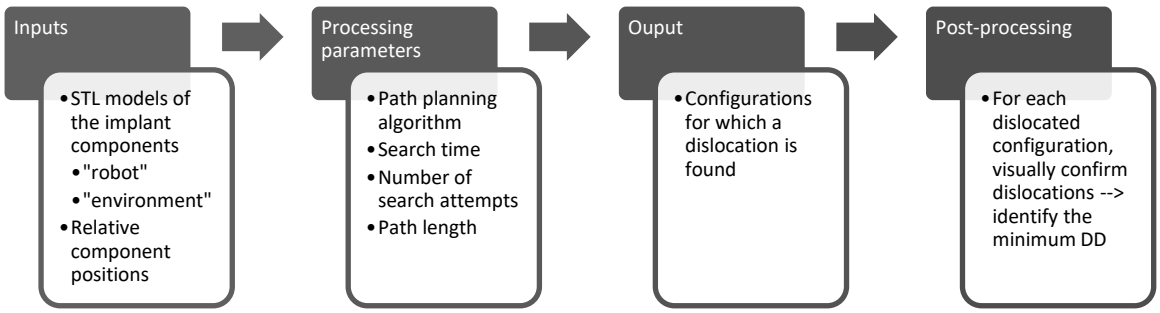
deviation of 1 mm (the accuracy to which a surgeon is able to reach clinically). The results also suggest that dislocation using the MATLAB code was easier than the mechanical rig – it is unsure which model was more accurate. Testing using the MATLAB dislocation model was computationally expensive, with assessment taking a total of 9546 seconds (~ 2.6 hours). The computational speed was also dependent on the number of points that require assessment. In this study the bearing surface area was reduced to a point cloud representation model based on the STL mesh. However, increasing the mesh density and hence the point cloud density may provide greater accuracy in the representation of the surface, this slows down the computational speed of the test.

Conclusions

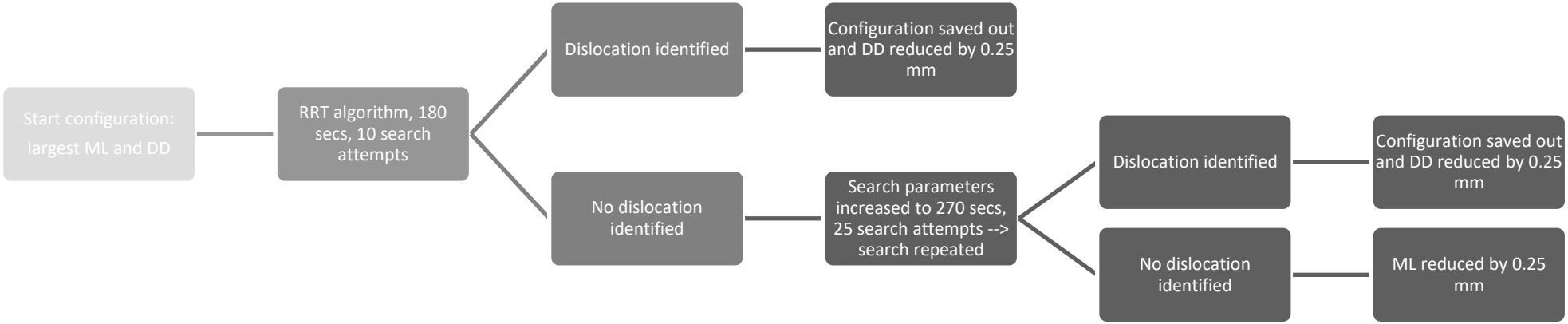
Automating the testing to allow improved efficiency is critical to enabling the development of a successful tool to quantify the risk of mobile bearing dislocation. This study successfully developed a computational dislocation assessment tool in MATLAB which is capable of receiving any new implant design, removes user dependency and is slightly more efficient than the mechanical rig. However, the tool makes many assumptions regarding bearing motion and most importantly, compared with the mechanical rig, the results obtained using the MATLAB dislocation tool have a different trend regarding the effect of ML translation on medial bearing dislocation and the results deviate more than 1 mm so a model that is more efficient and accurate was required.

APPENDIX B: ROBOTICS DISLOCATION ANALYSIS CODE

In the dislocation analysis tool, the RRT algorithm allows the bearing to identify a collision free path should one exist between the start (non-dislocated position) and the goal (dislocated position). If a collision free path is identified, the tools saves the configuration and these are later visually confirmed before identifying the DD for each mediolateral distance. The workflow is demonstrated below.



Appendix Figure 6. Flow chart to demonstrate how the dislocation analysis tool works.



Appendix Figure 7. The coding loop used in the dislocation analysis tool for assessing dislocations.

APPENDIX C: RELEVANT RESEARCH OUTPUT

Journal publications

I. Yang, G. Agustoni, D. Murray, S. Mellon (Submitted, under review), Mechanisms of mobile bearing dislocation in lateral unicompartmental knee replacement.

I. Yang, B. H. van Duren and H. G. Pandit (Submitted, under review), Lateral Unicompartmental Knee Replacement surgery for lateral knee osteoarthritis. Textbook chapter.

I. Yang, J. D. Gammell, D. Murray, S. Mellon (2022), Application of a robotics path planning algorithm to assess the risk of mobile bearing dislocation in lateral unicompartmental knee replacement. Sci Rep 12, 2068 (2022). https://doi.org/10.1038/s41598-022-05938-w

I. Yang, J. D. Gammell, D. Murray, S. Mellon (2021), The Oxford Domed Lateral Unicompartmental Knee Replacement implant: Increasing wall height reduces the risk of bearing dislocation. Proceedings of the Institution of Mechanical Engineers, Part H: Journal of Engineering in Medicine. doi:10.1177/09544119211048558

I. Yang, T. Hamilton, S. Mellon, D. Murray (2021), A systematic review on the dislocation of the mobile bearing in the Domed Lateral Oxford Knee: flat versus domed, vol 28, pp.214-28, The Knee.

I. Yang, J. D. Gammell, D. Murray, S. Mellon (2020), Application of a robotics path planning algorithm to assess the risk of mobile bearing dislocation in lateral unicompartmental knee replacement, CAOS 2020. The 20th Annual Meeting of the International Society for Computer Assisted Orthopaedic Surgery, EasyChair series EPiC Series in Health Science, vol 4, pp. 301-5.

J. Kennedy, H. Mohammad, I. Yang, S. Mellon, C. Dodd, H. Pandit, D. Murray (2020), Oxford Domed Lateral Unicompartmental Knee replacement: 10 year survival clinical outcome, Journal of Bone and Joint Surgery [Br], 102-B(8):1033-40.

Awards

- Doctoral Research Awards – 2nd place Engineering Sciences (**£200**), 2021
- Botnar Student Symposium – judges choice & audience choice, University of Oxford (**£100**), 2021
- DePuy Synthes Best Engineering PhD Project, Healthcare Technologies Student and Early Career Awards, Institute of Mechanical Engineers, London. (**£250**), 2021
- Oxford Medical Sciences Division Research Slam Winner (**£60**), 2020
- Oxford Travel Abroad Bursary (**£1,25K**), 2020
- EIT Health PhD Translational Research Fellowship (**€5K**), 2019
- Clarendon Scholarship - DPhil in Musculoskeletal Sciences, University of Oxford (3 years full funding), 2018

Conference presentations (podium presentation)

I. Yang, Jonathan D. Gammell, D. Murray, S. Mellon (Sept, 2021), Modifying the Design of The Oxford Domed Lateral Unicompartmental Knee Replacement Implant Reduces the Risk of Medial Bearing Dislocation. BORS conference 2021. Virtual.

Conference presentations (posters)

I. Yang, J. D. Gammell, S. Mellon, D. Murray (Feb, 2022), A novel computational dislocation analysis tool using robotics path planning algorithms for mobile bearing lateral unicompartmental knee replacement surgery, ORS conference. Tampa, FL, USA.

I. Yang, J. D. Gammell, S. Mellon, D. Murray (Feb, 2021), The Oxford Domed Lateral Implant: increasing tibial component wall height reduces the risk of medial dislocation of the mobile bearing, ORS conference. Virtual.

H. Mohammad, J. Kennedy, I. Yang, S. Mellon, H. Pandit, C. Dodd, D. Murray (Oct, 2020), The 10 Year Clinical Outcomes Of The Oxford Lateral Domed Unicompartmental Knee Replacement, EFORT conference. Virtual.

I. Yang, J. D. Gammell, S. Mellon, D. Murray (Oct, 2020), Application of a robotics path planning algorithm to assess the risk of mobile bearing dislocation in lateral unicompartmental knee replacement, EFORT conference. Virtual.

I. Yang, J. D. Gammell, S. Mellon, D. Murray (Mar, 2020), Application of a robotics path planning algorithm to assess the risk of mobile bearing dislocation in lateral unicompartmental knee replacement, STEM for Britain competition. Attlee Suite, Portcullis House, Parliament, UK.

I. Yang, S. Mellon, D. Murray (Sept, 2019), British Orthopaedic Society/Bone Research Society (BORS/BRS) conference, Dislocation of the mobile bearing in the Lateral Oxford Unicompartmental Knee Replacement (LOUKR): the effect of knee flexion, Cardiff, UK.

Internal talks/presentations

I. Yang, S. Mellon, D. Murray (presentation - Jun, 2021), Reducing Dislocation of the Mobile Bearing in the Oxford Domed Lateral Implant, Botnar Institute Student Symposium, Botnar Research Centre, NDORMS, Oxford, UK. Virtual.

I. Yang (presentation – Jul, 2020), S. Mellon, D. Murray, MSD Research SLAM, Using Robotics to Address the Problem of Mobile Bearing Dislocation in the Oxford Domed Lateral Implant. NDORMS, Oxford, UK. Virtual.

I. Yang (presentation – Sept, 2019), S. Mellon, D. Murray, IDEAL Collaboration Sandpit session, A New design for the lateral Oxford Unicompartmental Knee Replacement, Trinity College, Oxford, UK.

I. Yang (presentation - Jun, 2019), Global Scholars Symposium, Ethics in Surgical Implant Innovation, Rhodes House, Oxford, UK.

Yang (poster – Jul, 2019), S. Mellon, D. Murray, Medical Science Division DPhil Day, Dislocation of the mobile bearing in the Lateral Oxford Unicompartmental Knee Replacement (LOUKR): the effect of knee flexion, Tingewick Hall, Oxford, UK.